# OPERATOR PERFORMANCE AND LOCALIZED MUSCLE FATIGUE IN A SIMULATED SPACE VEHICLE CONTROL TASK

---

A Dissertation

Presented to

the Faculty of the Department of Psychology

University of Houston

---

In Partial Fulfillment

of the Requirements for the Degree

Doctor of Philosophy

---

By

James L. Lewis, Jr.

December, 1978

# OPERATOR PERFORMANCE AND LOCALIZED MUSCLE FATIGUE IN A SIMULATED SPACE VEHICLE CONTROL TASK

---

An Abstract of a Dissertation

Presented to

the Faculty of the Department of Psychology

University of Houston

---

In Partial Fulfillment

of the Requirements for the Degree

Doctor of Philosophy

---

By

James L. Lewis, Jr.

December, 1978

# ABSTRACT

Fourier transforms in a special purpose computer were utilized to obtain power spectral density functions from electromyograms of the _biceps brachii_, _triceps brachii_, _brachioradialis_, _flexor carpi ulnaris_, _brachialis_, and _pronator teres_ in eight subjects performing isometric tracking tasks in two directions utilizing a prototype spacecraft rotational hand controller. Analysis of these spectra in general purpose computers established general agreement with previous studies, aided in defining muscles involved in performing the task, and yielded a derived measure potentially useful in predicting task termination.

The _triceps_ was the only muscle to show significant differences in all possible tests for simple effects in both tasks and, overall, was the most consistently involved of the six muscles.

Monitoring of total average power in all muscles for contiguous sixteen-second intervals throughout each task provided consistent data useful in predicting task termination for all subjects. The total power monitored for _triceps_, _biceps_, and _brachialis_ dropped to minimal levels across all subjects earlier than for other muscles. However, smaller variances existed for the _biceps_, _brachioradialis_, _brachialis_, and _flexor carpi ulnaris_ muscles and could provide longer predictive times due to smaller standard deviations for a greater population range. The data for these muscles potentially provide information predictive of task termination up to approximately thirty-five seconds prior to the event.

This technique of relatively non-obtrusive monitoring and analysis has potential practical utility in work station and tool design, physical

training, medical applications, and extravehicular pressure-suited work activities for large scale space construction missions. Additional applied research must be conducted to identify derived measures sensitive to the needs of the designer and practically useful in design and real-time applications.

# ACKNOWLEDGEMENTS

This written expression of my appreciation and gratitude can only approximate the true feelings of indebtedness I feel to the many unnamed and the few named individuals who contributed to the successful completion of this effort. I offer my most sincere thanks to all of you.

To Dr. Sheer, my advisor, who taught my first experimental class long before we had the aid of computers and who provided help, counsel, and understanding when it was needed.

To my committee members, for serving, and for their helpful comments and encouragement.

To John Jackson and Earl LaFevers, for getting me started in the area.

To Dale Patenaude, Barbara Woolford, and Mike Thomas for their labors with the equipment and software.

A very special set of thanks to Jeri Brown and Bill Langdoc, whose interest in this study and whose labors in pursuit of those interests opened fresh avenues of thought and application.

But, most of all, to my family. My wife's parents, Sir Herbert and Lady Gamble, provided the peace and solitude of their home in Ireland, where much of the creative work was done. My parents, for so many years, have provided love and support--without their example, guidance and inspiration, this could not have been accomplished. My sons, Christopher and Philip, whose presence and happiness have so brightened my life, gave me added purpose. My wife, Ghislaine, who gave so much of herself to let me work, and who, most of all, always gave her love, was the inspiration I required. It is to them that I appreciatively dedicate this work.

TABLE OF CONTENTS

LIST OF TABLES

# LIST OF FIGURES

# CHAPTER I

## INTRODUCTION

The general purpose of this study was to determine if a relationship could be established between fatigue, as measured by electromyographic (EMG) recordings, and performance of a human operator in a simulated space vehicle control task. Specifically, the involvement of six arm muscles in a tracking task utilizing a spacecraft sidearm attitude controller was studied to:

(1) Ascertain conformity of the data to previous EMG studies of fatigue;

(2) Determine which muscle might best serve as an overall indicator of involvement in the task; and

(3) Determine if fatigue could be monitored in a manner that would allow prediction of performance degradation.

Fatigue, as a construct, is a complex phenomenon, and has been studied by psychologists, physiologists, neurologists, and many others. Bartley (1965) maintains that fatigue is a state which is measured by work decrement, and that more specific phraseology is required for adequate communication of fatigue-related research efforts. Ortengren (1975), quoting McFarland (1971), discusses various ways of categorizing fatigue: acute, caused by excessive use of an organ or body system; and chronic, arising from repeated acute fatigue. Alternatively some researchers prefer such subdivisions as local and general fatigue, the former arising from excessive use of a selected muscle and the latter associated with general tiredness and weakened motivation and concentration.

There is obviously no clear demarcation separating such arbitrary categories, but one aspect of the term that can be studied and quantified is that of localized muscle fatigue, a term suggested by Chaffin (1973). It is a condition reflecting physiological changes and measured by electrical activity in a muscle or local group of muscles brought about by repeated or extended use of such muscles.

Lippold et al. (1960) suggest that voluntary muscle contraction at a given sub-maximal level requires progressively greater effort to maintain, until a point is reached when the tension begins to fall. Maximum voluntary contraction continuously declines in tension across time. Therefore, whatever physiological events and processes are associated with the requirement to exert greater effort to maintain tension levels across time, the phenomenon may be arbitrarily referred to as muscle fatigue.

While definitions and applications vary widely across disciplines, the advent of electromyography has provided an objective, sensitive, and precise quantitative method of studying muscle activity. EMG signals are an indication of the electrical activity in muscles that arises whenever there is a voluntary or involuntary contraction of the muscle. Electrical activity associated with muscle changes occurs as a series of muscle action potentials (MAP) produced by motor units firing at certain frequencies. Electromyography has been studied qualitatively and quantitatively in both time and frequency domains since Adrian and Bronk

(1929), as cited by Komi and Viitasalo (1976), established a relationship between increased firing rates of motor units and increased tension of muscular contraction.

Prior to the advent of readily available digital processing capability, myoelectric signals were generally analyzed in the time domain. Developments in the 1950's eventually led to the application of digitally processed Fourier transforms to EMG data. This permitted study of complex EMG waveforms in the frequency domain and greatly extended research capabilities.

This study, then, utilizes Chaffin's concept of localized muscle fatigue as an arbitrary reference to the phenomenon described by Lippold et al. (1960), and applies spectral analytic techniques to investigate the myoelectric relationship of muscle activity to performance. In the event such a relationship can be established, what the phenomenon is arbitrarily named matters little. More important are the potential applications in control hardware and software design, in operational work-rest cycles, in extending the use of EMG techniques in manned spaceflight from research to operational applications, and in the potential increase in cost utility of time management techniques and reduced failures at the man-machine interface.

# CHAPTER II

## BACKGROUND AND STATUS OF EMG RESEARCH

### Time Domain Analysis of EMG

EMG analysis is currently accomplished in both time and frequency domains. Before advanced analog and digital processing capability became widely available, EMG research was confined to the time domain. Parametric analyses included such variables as amplitude, duration, motor unit discharge rate, and number of phases occurring in a period of interest. Recording methods were limited to analog output in the form of strip chart recorders, cathode-ray tube (CRT) presentations, magnetic tape and audio output.

Eason (1960) cites Seyffarth as having shown in 1940 that prolonged contraction of forearm muscles was accompanied by reported fatigue and discomfort, and by a reduction in action potentials. For the most part, however, little quantitative analysis was possible with such methods.

As analog data gathering and analysis techniques improved, researchers began to evaluate recorded EMG by various integration methods to better ascertain relative work loads associated with voluntary muscle contractions. A frequency counting system was developed by deVries (1968a) in which the count registered per unit time was related to EMG voltage. He used this system (1968b) to conduct a series of experiments for developing a math model for fatigue curves. Both isometric and isotonic exercises were tested across various percentages of maximum voluntary contraction (MVC). He concluded that a generalized model was unlikely to be determined, at that time, which could describe the relationship of various parameters entering into muscle fatigue; he did, however, demonstrate a linear relationship between integrated EMG (IEMG) and various mid-range percentages of MVC used in the study. These results were consistent with earlier studies by Lippold (1952) and by Edwards and Lippold (1956) who utilized similar recording and analysis techniques. The Lippold (1952) results were determined to be consistent regardless of electrode placement (as long as the muscle belly or tendons were used), proximity, or skin resistance. Currier (1970) also utilized a frequency counting method to verify a linear relationship between the progressive increase of electrical activity from a sustained static contraction of a single muscle and the time of contraction.

Scherrer and Bourguinon (1959) found that IEMG increases with time for constant or increased work levels. They also demonstrated increased action potential amplitudes and decreases in the frequencies at which the higher amplitudes occurred with time.

In a study of the effects of magnitude and duration of sustained isometric muscular contractions voluntarily maintained until exhaustion, Eason (1960) found a gradual increase with surface IEMG and time. As an explanation, he suggested that additional motor units were progressively recruited to compensate for the loss in contractibility due to impairment, and that the action potentials of the new units summed with those already active units to more than offset the amplitude decrement of the latter.

Utilizing IEMG analysis techniques, Lippold et al. (1960) found that, as voluntary effort to maintain a given tension level continued, electrical activity, amplitude, and synchronization of motor unit firing all increased, and action potential spikes moved closer together. These findings held true for both surface and needle electrodes. Migration of activity from one muscle to another was also demonstrated. Electrical activity in a primary contributing muscle had ceased after five minutes in an isometric task, but performance had not degraded. Through the use of needle electrodes, transfer of the function to a deeper muscle was traced.

Chapman and Troup (1969) demonstrated a linear relationship between IEMG and the external force produced by a single muscle under isometric conditions, and found no change in this linear relationship as a function of gains in strength.

Zuniga and Simons (1969), however, reported a non-linear (quadratic) increase in average EMG voltage with increased tensions up to maximal levels, although their results appeared to indicate linearity existed at tension levels below approximately 70% MVC.

In tests of a single leg muscle, Kuroda et al. (1970) demonstrated a linear relationship between IEMG and force for almost all of the sub-maximal range. As force closely approached MVC, the EMG activity increased in a manner best approximated by an exponential function.

Bouisset and Maton (1972) found high correlations and a linear relationship between surface IEMG and intramuscular (inserted wire electrodes) IEMG for the <u>biceps brachii</u> muscle in a dynamic anisometric-anisotonic sub-maximal contraction task.

## Frequency Domain Analysis of EMG

Once digital computer power was available to research efforts, Fourier analysis could be applied to complex EMG waveforms to obtain a very high frequency resolution for a sequence of time intervals (Broman, 1973). The Fast Fourier Transform enables analysis of a complicated signal in terms of its component parts, expressed in terms of an infinite number of sinusoids of individual amplitudes, frequencies, and phases. The power spectrum of a signal, then, is a description of how the total power, averaged for a selected time interval, is distributed in the frequency domain.

Ortengren (1975) quotes Richardson (1951) as the first reporting researcher utilizing power spectral changes in EMG studies, but another decade was to pass before power spectral density (PSD) studies were reported on a more frequent basis.

In an early medical application of Fourier analysis to EMG, Cenkovich and Gersten (1963) developed an index of harmonic spectra to compare the high frequency extent of major amplitude harmonics and the magnitude of the slowly decreasing residual harmonics in order to differentiate pathologically short potentials from high frequency peaks of normal potentials.

Chaffin (1969a) used an analog method (filter bank analyzers) to evaluate power spectral differences in myopathic and neuropathic versus normal muscles. He found significant shifts of the EMG frequency spectra toward higher frequencies for the pathological muscles, and an increase in low frequency component amplitudes relative to high frequency components for asymptomatic individuals.

Using digital analysis, Chaffin (1969b) tested a single muscle to exhaustion in isometric tasks and related reported subject discomfort, decreased eye-hand coordination, and decreased hand steadiness to shifts in EMG power from high to low frequency bands. He found that fatigue (as determined by power increases in low frequency bands) increased at an accelerating rate as a function of increased load. There was a linear decrease in power for the higher Hz bands. In this study, he considered total power to reside over a range of four to 200 hertz, and utilized 4-30 Hz and 60-100 Hz for the low and high frequency bands for the analysis of power shifts.

In an attempt to relate combined anatomically and physiologically based concepts of skeletal muscle actions to engineering principles associated with design and analysis of man-machine interfaces, Chaffin (1969c) defined functional states of subjectively reported discomfort for shifts in EMG power from the 40-70 Hz band and related the subjectively reported categories to percent increase in eye-hand coordination test times.

The specific frequencies examined vary across studies and researchers by a considerable amount. Chaffin (1973) suggested that a practical measure of frequency shift could be found in the region from above 70 Hz to below 40 Hz.

Lloyd (1971) also related power shifts to subjectively reported pain and, in one of the more quantitative approaches to band selection, reported the dominant amplitudes at the maximum reported pain levels to lie between 12 and 50 Hz.

Kwatny et al. (1970) studied frequency shifts (over a range of 0-400 Hz) of two arm muscles before and during fatigue for two levels of contraction and found that fatigue was characterized by more power in the lower half of the spectrum, while signals produced before fatigue had greater power in the upper half of the spectrum. They also demonstrated that relative average power increased during fatigue, and attributed this to an increase in the number of muscle fibers discharging synchronously.

In an extension of previous studies, Johansson (1970) used four octave filters with center frequencies from 50 to 1600 Hz to demonstrate a relative high frequency decay, which was reversed to a high frequency increase during recovery.

There have been few studies relating fatigue-EMG to performance criteria other than maintenance of a selected force level. One example of such a study is that by Lance and Chaffin (1971) where the relationship was studied between EMG changes and reaction time, initial adjustment time, overshoot values, stabilizing time, and movement time in a simple arm movement pointing task. They found reaction time and initial adjustment time were not significantly affected by fatigue, while overshoot, stabilizing time, and total movement time were significantly affected due to fatigue.

Komi and Viitasalo (1975) used both IEMG and spectral analysis to study voluntary contractions up to maximal levels. Their studies established that IEMG increased quadratically with muscle tension, and, while spectral analytic data were not studied for linearity characteristics, they found that amplitude-rise time, number of spikes, and amplitude-rise time ratio for the averaged motor unit potential all increased as muscle tension increased.

In a study which compared various derived measures, Viitasalo and Komi (1975) studied the reproducibility of EMG results for the IEMG, Averaged Motor Unit Potential (AMUP), power spectrum, and Mean Power Frequency (MPF), and found that IEMG and AMUP demonstrated better within test session reproducibility than across days, while power spectral analysis and MPF were more reliable across days.

## Fatigue Measured by EMG and Work Decrement

The relationship of electromyographically measured fatigue and subjectively reported pain had been well established by the early 1970's, as was its relationship to laboratory pointing and coordination studies (Chaffin, 1969b, and Lance and Chaffin, 1971). At this point, various researchers began to orient their efforts toward applied studies which more closely approximated or were an actual part of the work environment.

LaFevers (1974a, b) utilized surface electrodes on three muscles in a push-pull task in shirtsleeves and a pressurized space suit, and demonstrated significant suit effects over shirtsleeve fatigue responses. In some instances, the pressurized space suit aided the work task and in others brought on fatigue more rapidly in relation to the shirtsleeve condition. He also found differential responses across muscles and a differential sensitivity of muscles to various reach positions.

In an orbital zero-gravity space flight study, LaFevers et al. (1975) demonstrated that two antigravity leg muscles showed heightened susceptibilities to fatigue as a function of weightlessness. These findings were evidenced by differentially significant power spectral density increases in high frequency bands similar to responses of pathologically diseased muscles.

A study of automobile assembly line workers by Ortengren et al. (1975) established that heavy body polishing work produced statistically significant higher intensities of fatigue incidents than light work, and that the distribution of intensity values over muscles corresponded to the engagement of the muscles in the work task.

Then, in a study of equally trained but inexperienced versus experienced welders, Kadefors et al. (1976) demonstrated EMG measured fatigue differences between the two groups. Experienced welders showed localized muscle fatigue in the supraspinatus muscle only, whereas inexperienced welders showed fatigue in three shoulder muscles.

## Statement of the Problem

While these applied studies have provided invaluable insight into man-machine interface design requirements and training methods, there remains an unexplored area of predicting performance decrement by EMG monitoring of muscle activity. Additionally, a definition of the involvement of specific muscles or muscle groups in a space vehicle control task would greatly aid in hand controller design efforts.

The previously discussed research and study efforts have confirmed shifts of the power spectrum toward lower frequencies with the onset of fatigue caused by sustained muscular contraction, and linearity in the percent change with time has been basically confirmed in the mid-ranges of MVC. The effects of skin resistance and capacitance coupling on EMG amplitude and power spectra were studied by Schanne and Chaffin (1970), who determined that differences in electrode diameters, electrode pastes, and EMG measurement procedures, in general, probably do not exclude direct comparison of EMG spectrum results.

Electromyography and spectral analysis can therefore, be reasonably applied to identify those muscles involved in the utilization of a spacecraft sidearm controller and to investigate the potential predictive aspects of the PSD/fatigue relationship for performance decrements.

# CHAPTER III

## METHOD

The system utilized consisted of one special and three general purpose digital computers with associated peripherals for handling simulation requirements, data input, recording and analysis, and various forms of output. The complete system is diagrammed in Figure 1.

The Hewlett-Packard (HP) 2100A computer served multiple roles by controlling the tracking task simulation, recording error and physiological data occurring as a function of the simulation, and by serving in the data reduction mode in early stages of analysis.

A symbol generator, a Lear variable-configuration three-axis rotational hand controller (RHC), shown in Figure 2, a Tektronix 610B cathode ray tube (CRT) on which the symbol was displayed, analog-to-digital converters, and various power supplies comprised the tracking task/symbol display and control part of the system. The RHC configuration was variable with respect to pivot points, displacements, force levels, and electrical gains, and was mounted on a stand which could be adjusted for seat and arm height, arm length, and forearm inclination.

During the tracking task, symbol movement on the CRT was programmed for and controlled by the HP 2100A which also processed programs that interpreted and recorded the digitized hand controller data input by the subjects to track the programmed symbol movement. In addition, the HP 2100A provided a record of performance quality by recording, via the HP 7970B tape drive, error data between the programmed symbol position and

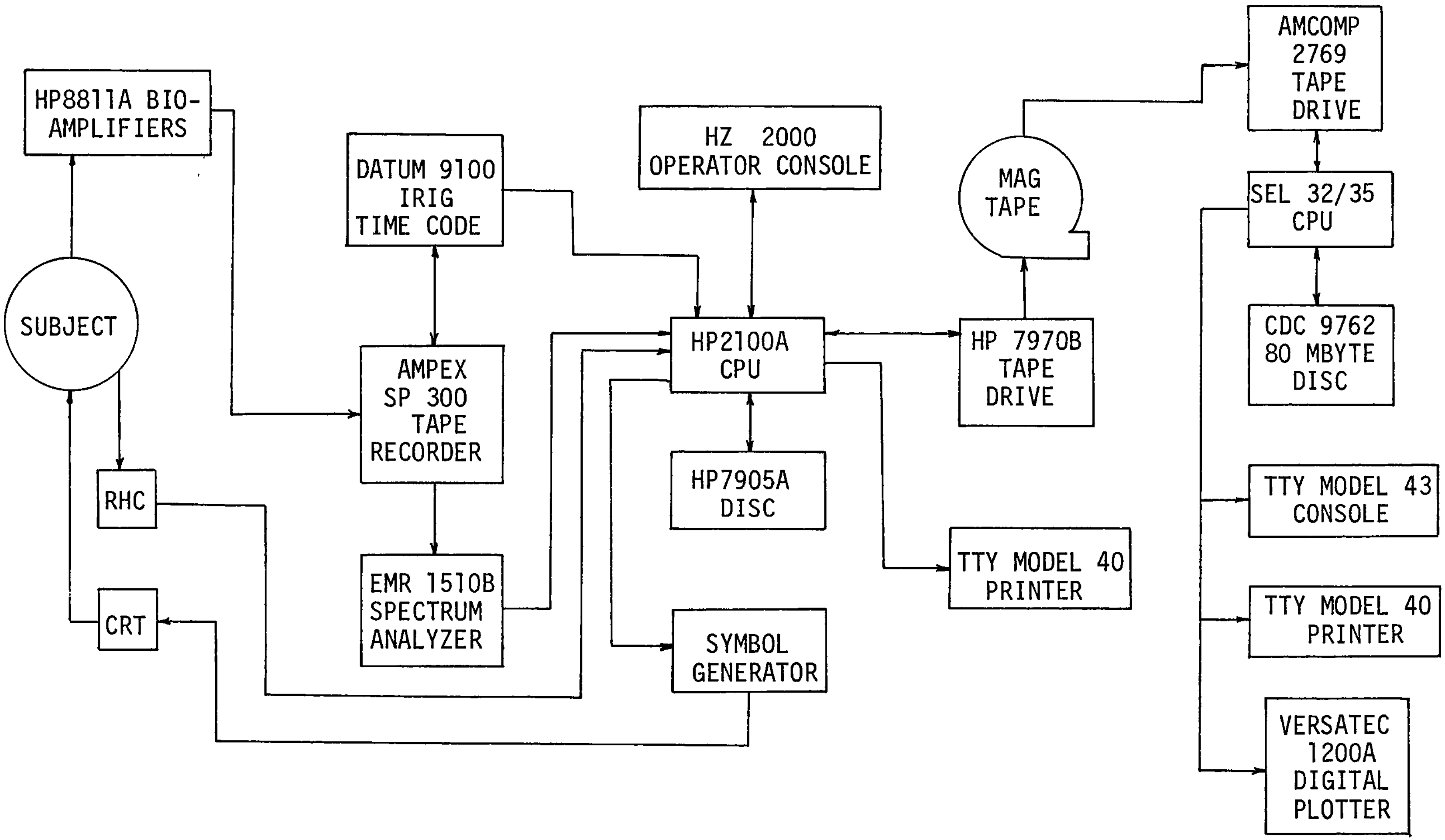



Figure 1. System Block Diagram

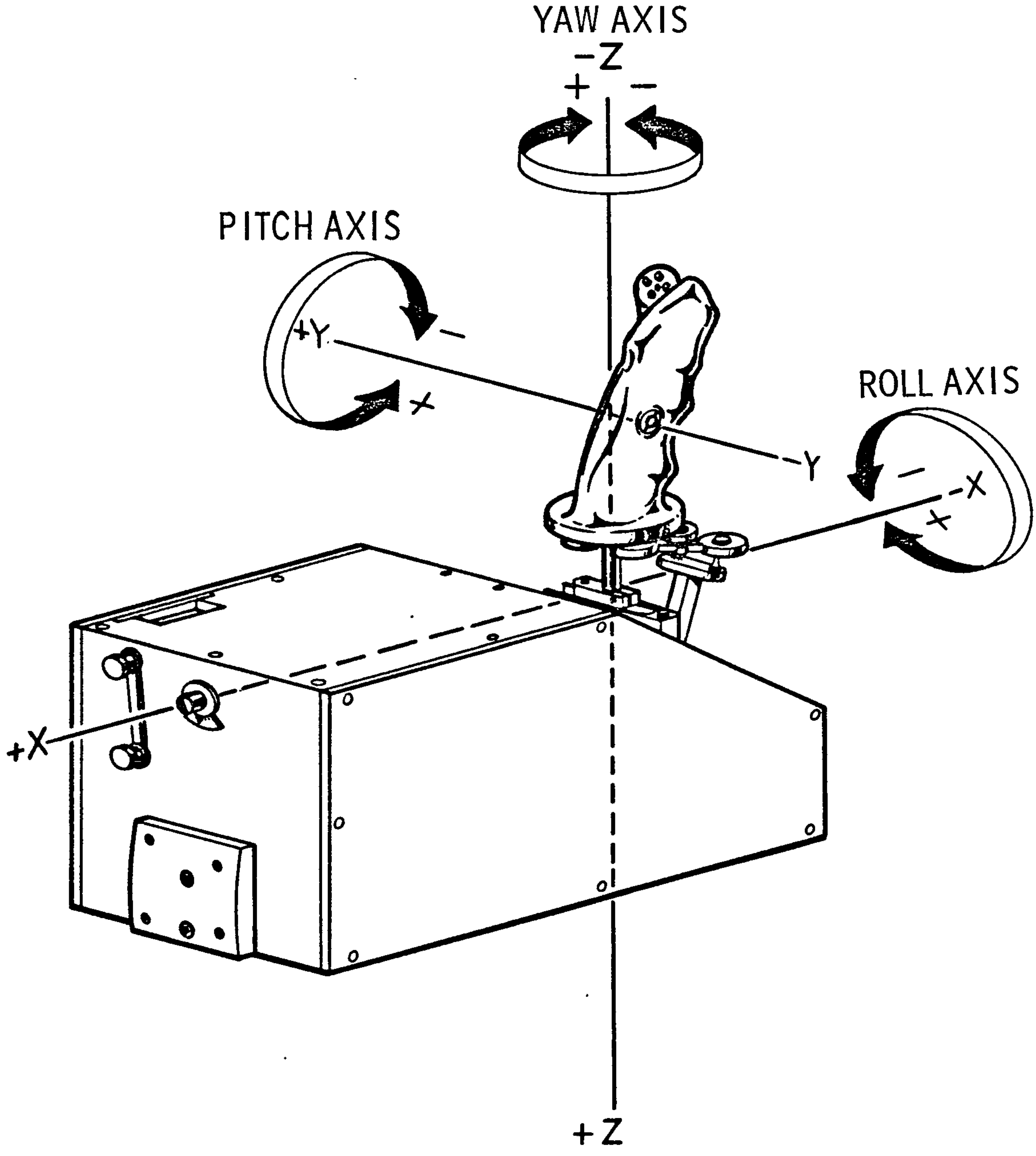



Figure 2. Lear Rotational Hand Controller

the tracking position commanded by the subjects. Programs which read the spectral information from the Schlumberger Electromyographic Recorder (EMR) 1510B were executed on the HP 2100A, and the resulting data files were stored on the HP 7905A disc for statistical analysis.

The physiological data collection and spectral analysis subsystem consisted of the EMR 1510B Spectrum Analyzer, six HP 8811A bioelectric amplifiers, Beckman 11 mm skin electrodes, an Ampex SP 300 seven-channel tape recorder, and a Datum 9100 time code generator. For testing and data recording, six channels of EMG signals were amplified by the HP 8811A's and placed on the SP 300 tape at 7-1/4 inches per second in the FM mode along with an IRIG time code generated by the Datum 9100. The same time code was recorded simultaneously with tracking task error data on the HP 7970B digital tape. In the first phase of analysis, a single channel of physiological data was played back from the SP 300 through the EMR 1510B analyzer for spectral analysis, and a disc file created for the data on the HP 7905A disc. In the translate mode, the Datum 9100 decoded the time signal and placed it in the disc file along with relevant spectral data. Finally, the digital error data tape and tapes of the HP 7905B physiological data files were used on the Amcomp 2769 tape drive with appropriate software to establish disc files on the Control Data Corporation (CDC) 9762 disc for subsequent analysis using the Systems Engineering Laboratories (SEL) 32/35 central processing unit (CPU).

## Subjects

The subjects for this experiment were four male and four female aerospace engineers, professional level administrators, and test subject pool personnel who had previously participated in various studies and simulation efforts. They ranged in age from 24 to 41 years, were of average to good physical fitness, and while five were qualified pilots, all had aerospace simulation experience with sidearm hand controlled tracking tasks. All were familiarized with the test apparatus and tracking task prior to data collection.

## Procedure

Based on electrode testing accomplished by Geddes and Baker (1968) and Geddes (1972), silver-silver chloride skin electrodes were used for EMG potential sensing. Beckman Instruments Company units were selected for equipment commonality reasons. The 11 mm electrode surface was recessed approximately 2.75 mm in a 16 mm diameter molded plastic housing. Prior to use, the electrodes were cleaned and soaked in a mild saline solution (1% salt in distilled water), after which they were dried and Beckman adhesive collars placed over the mating surface. Beckman electrode electrolyte was then used to fill the cavity. The skin surface area over the selected muscle sites was prepared by vigorous rubbing with cheesecloth saturated with 70% isopropyl alcohol. Muscles were palpated according to instructions suggested by Hinson (1977). The electrode pairs were then placed approximately 2.1 cm apart over the belly of each muscle that could be palpated. The *brachialis* could generally not be palpated, so the electrodes were placed in the area where the muscle would surface if developed to that point. All sites were located by reference to the *Stereoscopic Atlas of Human Anatomy* (Bassett, 1960). Electrodes were "aged" according to the procedure recommended by Schanne and Chaffin (1970) for 30 minutes or more and checked to ascertain interelectrode and electrode-to-ground resistances were less than 20,000 ohms.

Each electrode was then connected to the appropriate bioelectric amplifier and representative trial runs conducted to set gains and check for system noise. Since the frequency range of interest in this study was 12-240 Hz and the primary noise, generated from the 120 Volt, 60 Hz facility power supply, was within this range, the data analysis program was modified to control for this component and its harmonics. The remaining noise was flat to within approximately ±4 db across the frequency band of interest and of insignificant magnitude.

Subjects were seated in the fixture and briefed on the tracking task. Seat and arm heights were adjusted for consistent viewing and RHC actuation angles, respectively.

## Task

Symbol movement dynamics were modeled after an early version of the Manned Maneuvering Unit (MMU)--a backpack propulsion unit for locomotion and attitude control in free space to be used by a crewmember in a pressurized space suit. The effect of pressurized space suits on mobility and strength varies according to anthropometric fit, suit design and the nature of the task. For this reason, it was determined that the initial study effort of any task involving EMG analysis should first be accomplished without space suit encumberance to ascertain the nature of any fatigue-predictive capabilities.

The digital model of the MMU accommodated control system dynamics, mass properties and pilot field-of-view. The control system modes include acceleration command with attitude hold and rate command capabilities. For purposes of this study, only the rate command system was used. The RHC was spring loaded to a center detent position in the roll, pitch, and yaw axes. Movement of the RHC out of the electrical deadband defining the detent position in any half-axis simulated an MMU thruster firing which resulted in movement. This movement was simulated by changing position of the symbol on the CRT screen. In addition, symbol movement on the screen was independently pre-programmed to simulate the movement of an object (target) in space. Thus, the tracking task involved movement of the RHC in the direction of pre-programmed symbol movement to allow the pilot's line-of-sight to be maintained as a constant relative to the target. The effect of such action was simulated on the CRT screen by the target symbol remaining beneath a fixed reticle when RHC movement exactly matched the target pre-programmed maneuver.

The pilot lateral field-of-view was modeled by programming the CRT screen width to be approximately 190°.

The term "rate command" is used in this application to mean that once the RHC was out of detent and causing a thruster firing, the MMU/man system accelerated at 10°/second$^2$ until either the commanded rate or a maximum rate of 20°/second was achieved. Hand controller displacement was proportional to the commanded rate, i.e., maximum displacement in any one direction commanded a rate of 20°/second. A given commanded rate was maintained as long as the RHC was held in the same position. Once the RHC position was changed, the rates changed again at 10°/second$^2$ until the newly commanded rate was achieved.

Error was calculated by subtracting the achieved rate from the programmed target rate. This term was calculated every 100 milliseconds, averaged for one second and output to disc file in one-second increments, along with the IRIG time from the Datum 9100.

In an earlier study by Moore (1977) to define the muscles involved in the control task, some difficulty was encountered because the maneuver simulated was dynamic with respect to muscle involvement and the forces utilized in the RHC were low. To alleviate those problems, several changes were made in the task and equipment. It was first decided to study each half axis separately and to redesign the task such that it required constant isometric contraction to exhaustion. Due to the volume of data resulting from these decisions, only two half axes could be included in this effort; the "roll left" and "roll right" directions (see Figure 2) were chosen because of the ease of implementation in the Lear RHC. The eight subjects were then checked for maximum voluntary contraction (MVC) in these directions, and springs selected for the RHC to match one-third of the average MVC of the group in each direction when the RHC was displaced at 95% of its maximum travel. Rather than using a fixed length task, the tracking simulation was run until the subject could no longer move the RHC out of detent against the spring.

Each half axis task consisted of an initial two minute period of no command during which EMG data were taken to provide confirmation that all parts of the system were operative. At 120 seconds, the pre-programmed target maneuver caused the symbol to accelerate at $10°/\text{second}^2$ for 2 seconds to a rate of 9.5°/second. This rate was programmed as constant for 20 minutes. No subject maintained the voluntary contraction longer than 16 minutes.

The symbol representing the target was a circle approximately 1.5 cm in diameter. The pre-programmed maneuver caused the circle to move (in the absence of an RHC command) left from the center toward the edge of the CRT screen for the roll left maneuver, and to move right for the roll right maneuver.

The order of presentation of the two half axes was counterbalanced across subjects. A rest period between half axis tasks of 30 minutes was used in all cases.

## Data Preparation for Statistical Analysis

Because the EMR 1510 has a single channel digitizer, each tape, after completion of a given run, was played back through the spectrum analyzer six times (once per muscle). Therefore, for the eight subjects, each making one run to the left and one to the right, 96 passes of varying lengths through the EMR were made. Each spectral analysis pass took 1024 samples per second, averaged these data points for each second, and then averaged 16 successive one-second samples to compute the amount of power existing at each frequency value between 12 and 240. Contiguous 16-second, averaged data samples were made for each muscle from beginning to completion of the run. Thus, a total of 12-1/4 hours or 45,158,400 samples of physiological data per frequency value were reduced in the spectral analysis process for the eight subjects.

These spectral data were then stored in a disc file for subsequent analysis. The program which transferred the power spectra to the HP 7905A disc also linearized and normalized the data. Normalization was based on the largest value for a given muscle in an entire run.

For the purpose of studying power shifts, intervals derived from Lloyd (1971) were selected. The 12-50 Hz interval was specified as the low frequency band, and 62-100 Hz range as the high frequency band of interest. The percent spectral power contained in these bands was used in the analyses of variance (ANOVA's) discussed in Chapter IV, in the calculation of a spectral index (SI) suggested by Moore (1977), and in the program used to plot the percent of total power (XI) in each band as a function of time. Examples of the SI and XI graphs are shown as Figures A-1 and A-2, respectively, in Appendix A.

Power was summed across all frequencies from 12 to 240 Hz to yield a total power (TP) measure for each 16-second analysis interval. A sample plot of TP appears in Appendix A as Figure A-3. The magnitude and frequencies at which the maximum and minimum powers occurred in each 16-second analysis interval were also examined. Examples appear as Figures A-4 and A-5 in Appendix A.

Figures A-6 and A-7 show plot samples of the difference in XI for the high and low frequency bands. Figures A-8 through A-11 demonstrate frequency/amplitude plots for contiguous 16-second intervals, while Figures A-12 through A-15 show frequency/amplitude plots for contiguous 16-second intervals averaged over the following four periods: From task start to the interval prior to error increasing to greater than 5°/second; from the end of the previous period to the end of the interval in which error exceeded 5°/second; from the end of the previous period to the start of the last full interval prior to task termination; and the final full 16-second interval prior to, but not including, task termination.

Separate programs transferred the error and physiological data disc files from the HP 7905A to digital magnetic tape using the HP 7970B tape drive. This tape was then read from the Amcomp 2769 tape drive, and the data placed in file on the CDC 9762 disc where the SEL 32/35 CPU processed the programs for plotting the simulated three-dimensional power spectra plots shown in Appendix A as Figures A-8 and A-11, respectively. While the capability of the SEL system was required in these applications, the HP system accomplished all other data analyses discussed in this study.

All plots in Appendix A were formatted in the process of seeking a reliable event or set of events to aid in predicting the onset of performance degradation or task termination. While the data presented in the appendix were consistent with previous findings, the information so formatted did not satisfy this primary intent and is, therefore, presented in sample form for information purposes only.

# CHAPTER IV

## RESULTS

In order to establish baseline performance conditions and relate these to electroymyographically manifested fatigue, the combination of an isometric task and two tracking directions was selected and six muscles monitored for PSD changes in low and high frequency bands. The *biceps brachii* (BB), *triceps brachii* (T), *brachioradialis* (BR), *flexor carpi ulnaris* (FCU), *brachialis* (B), and *pronator teres* (PT) were selected based on their involvement in either or both tasks (Basmajian, 1967; Hinson, 1977; Kendall et al. 1971; Wells et al. 1976).

The spectral index (SI) investigated by Moore (1977) was derived by dividing the power in the 12-50 Hz band by the power in the 62-100 Hz band. The SI performed as expected--i.e., increasing with time for a given isometric task--but was not relatable to those manifestations of performance decrement investigated in this study.

The percent of total power (XI) in each band also provided expected results in that XI in the low band increased with time for the isometric task while XI decreased in the high band. For each of the six muscles across all subjects, the low band initially contained a smaller percentage of total power than at task end, with this condition being reversed for the high band.

Total Power (TP) was derived by summing the normalized power for each frequency value from 12-240 Hz for each contiguous 16-second time interval. TP rose rapidly with the beginning of the task, peaked, and declined as fatigue increased.

The normalized maximum and minimum powers between 12 and 240 Hz (MI) were recorded for each contiguous 16-second interval. The minimum power was basically flat throughout the task interval. The plotted value of the maximum power across time generally appeared to approximate the shape of the TP curve.

The frequencies at which the normalized maximum and minimum powers occurred for each contiguous 16-second interval (FR) were also recorded. Minimum powers generally occurred at frequencies above 200 Hz while the maximum powers occurred below 100 Hz with the frequencies of the two bands diverging as the duration of the isometric task increased.

Of these derived measures, only TP could be related in a consistent and meaningful manner to tracking performance.

The first 120 seconds of every run formed a quiescent period during which system operation was verified and no muscle activity was present. Absence of muscle activity can be observed in the TP and MI plots as they are representative of actual power. Because SI and XI are ratios, the plotted values for the initial 120 seconds are not null or near null values. No data from this quiescent time period were utilized in any of the analyses.

For the sample plots shown in Appendix A, the tracking error was greater than 5°/second after 248 seconds run time; run termination (the point at which the subject could no longer overcome the initial breakout force of the hand controller spring) occurred at 338 seconds.

Because SI, XI, MI, and FR did not appear to have any discrete characteristics that could be used as predictors of performance decrement, these measures were set aside for future consideration. As each did, however, behave in a predictable manner, it is likely that use of different measurement sensitivities, test/task designs, and criteria could find them useful.

Figures 3-6 present the average percent EMG power in each of the six muscles investigated as a function of the following three discrete times: Start of the task; the point at which tracking error became greater than 5°/second; and task end. Four separate conditions are plotted: Roll left, low and high frequency bands; and roll right, low and high frequency bands.

Because the primary intent of the study was to identify the muscle or muscles which could best be utilized in these tasks as an indicator of fatigue and a predictor of performance decrement, each muscle and task was separately analyzed.

A repeated measures design (Edwards, 1960) was developed to take advantage of the relatively low number of subjects available and the volume of data resulting from the tests. The percent power in each of the frequency bands was calculated based on the total power in the 12-240 Hz band for each 16-second interval and was used as the dependent variable. The time intervals of interest were defined as the first full interval of muscle electrical activity after the task beginning (Start), the interval in which the error rate built to a sustained value of greater than 5°/second (E>5), and the last full 16-second interval prior to task termination (End).

While the ANOVA is a relatively robust test with respect to homogeneity of variance, it is useful to understand in what direction lies the

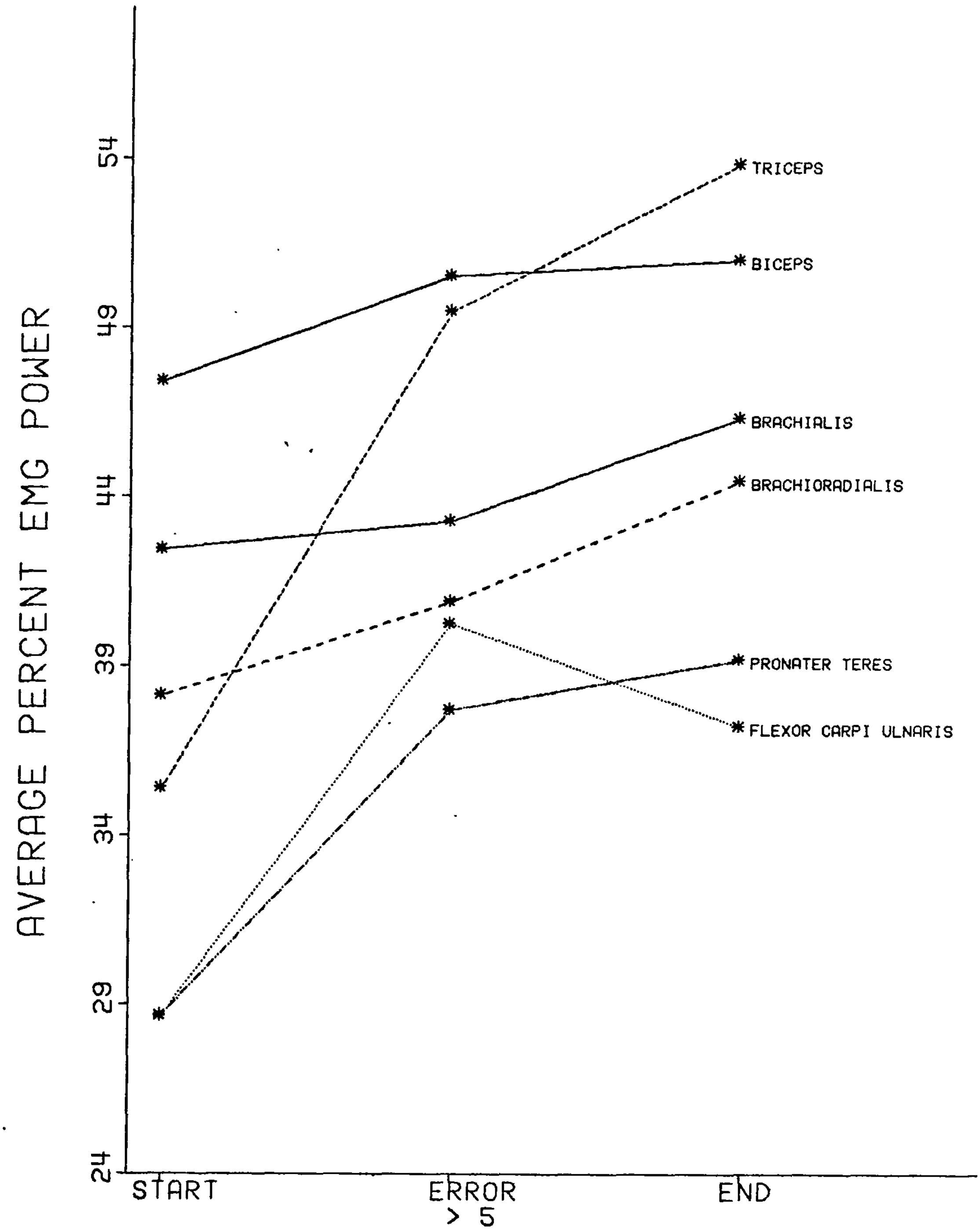



FIGURE 3

MAGNITUDE OF EMG PERCENT POWER SHIFT FOR 12 - 50 HERTZ BAND ROLL LEFT

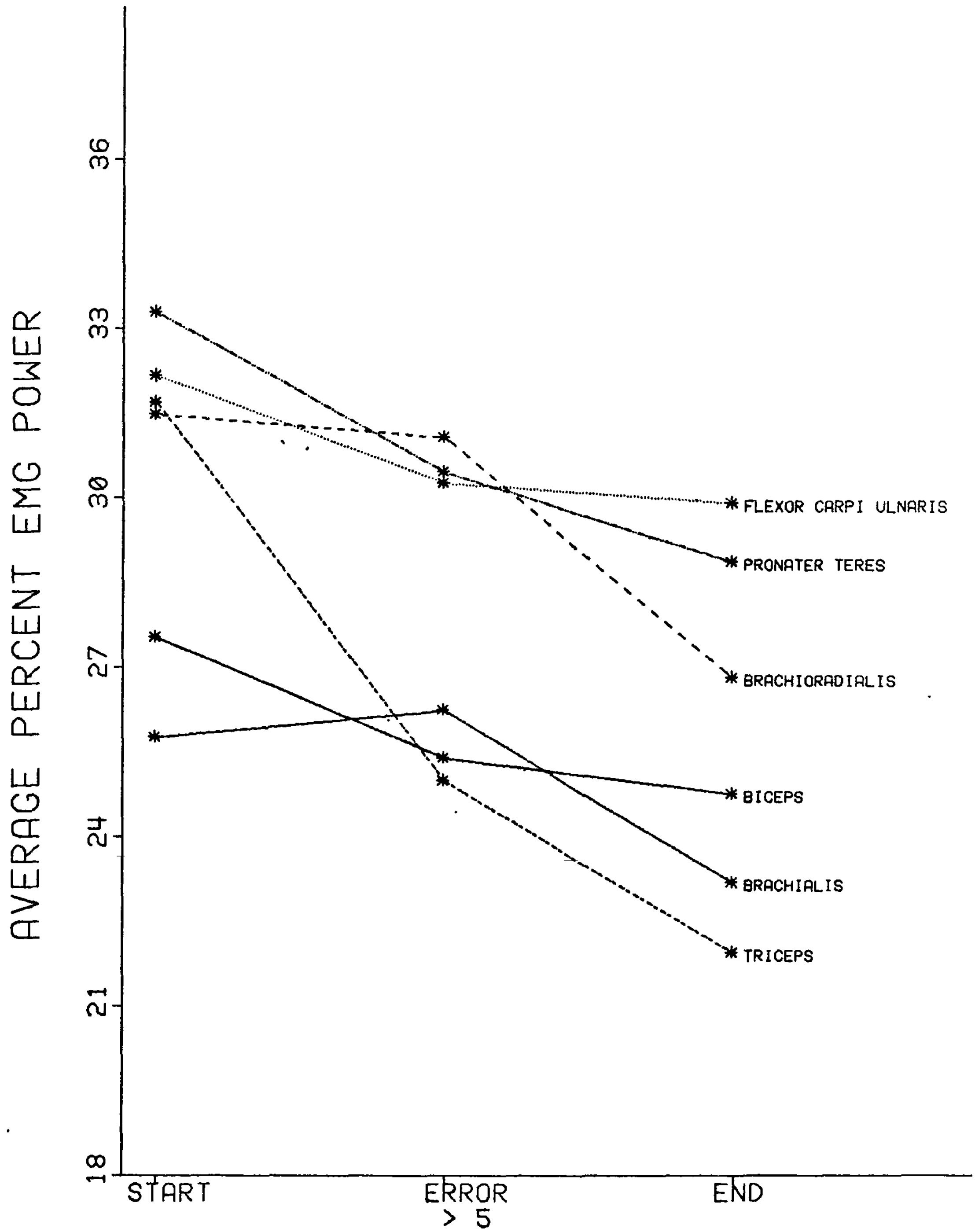



FIGURE 4

MAGNITUDE OF EMG PERCENT POWER SHIFT FOR
62 - 100 HERTZ BAND
ROLL LEFT

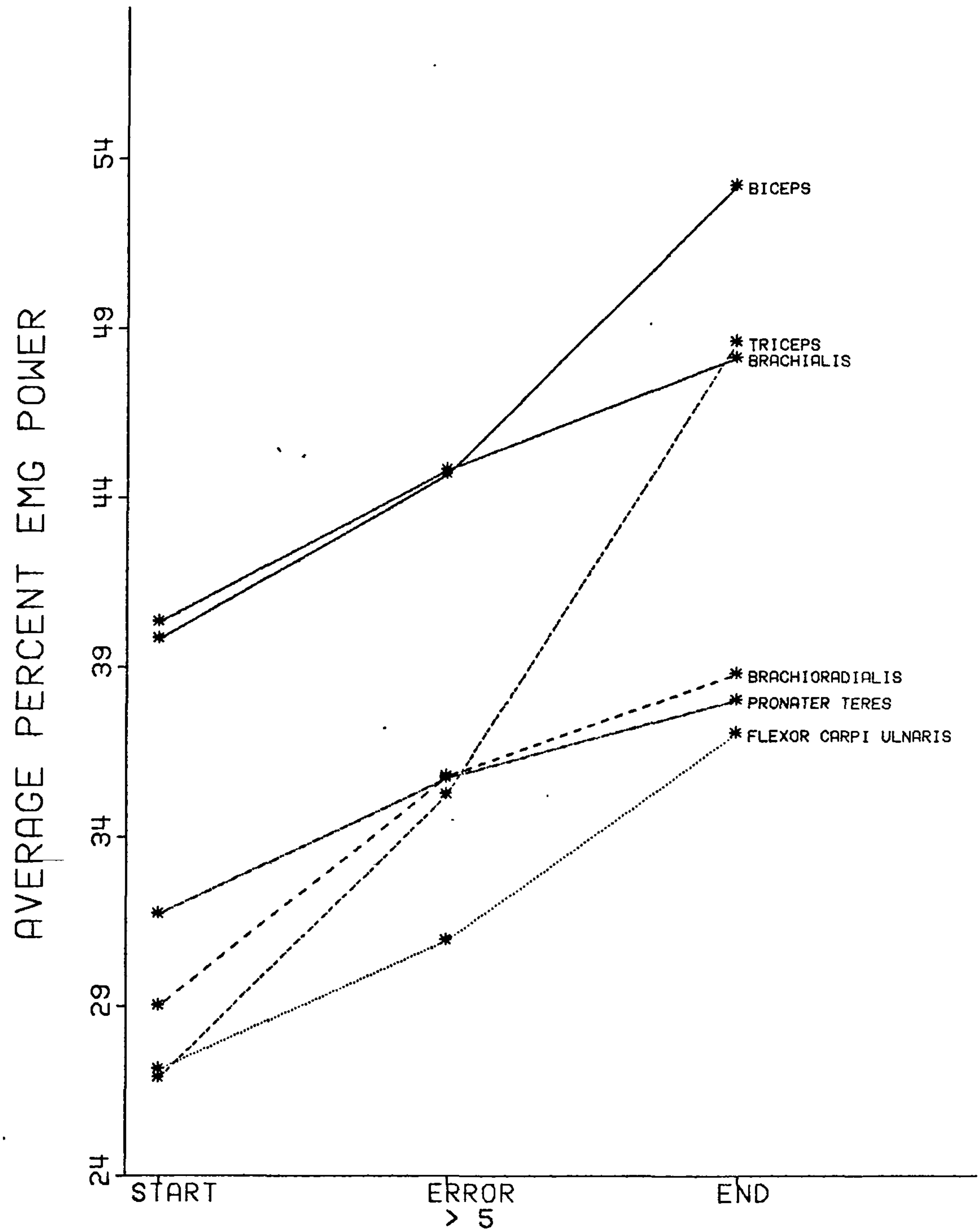



FIGURE 5

MAGNITUDE OF EMG PERCENT POWER SHIFT FOR

12 - 50 HERTZ BAND

ROLL RIGHT

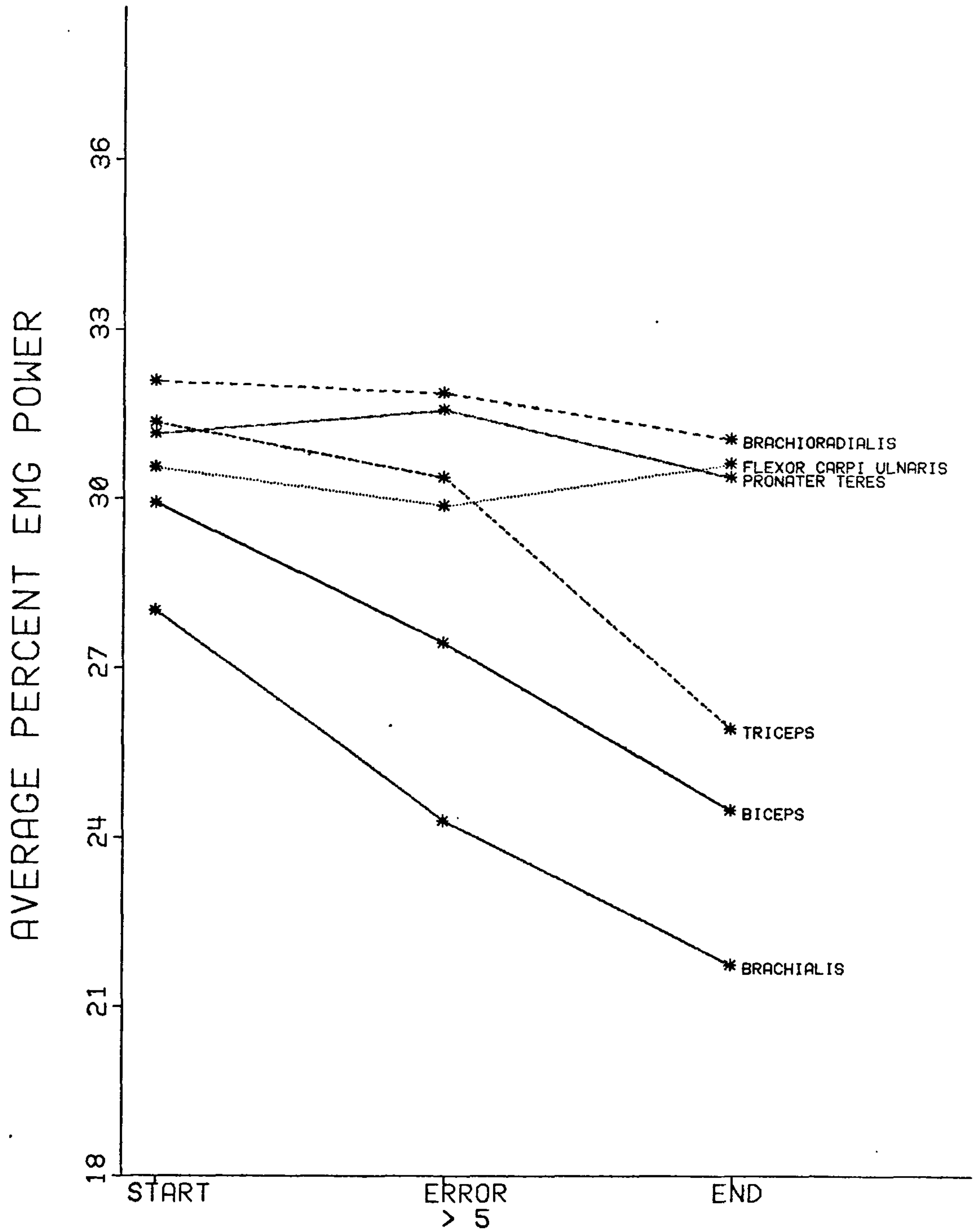



FIGURE 6

MAGNITUDE OF EMG PERCENT POWER SHIFT FOR
62 - 100 HERTZ BAND
ROLL RIGHT

likelihood of making an error. Therefore, Hartley's (1950) F-Maximum test was applied to the data to check for subject homogeneity across each muscle. The results of these tests are shown in Table 1.

The data were treated with the arcsin transform recommended for proportional data by Winer (1962) and analyzed in the treatment by treatment by subjects design developed in the computational text by Bruning and Kintz (1968). The Newman-Keuls tests for ordered means recommended by Winer were used to test for simple effects where statistically significant results were obtained. The ANOVA results and the tests for simple effects appear in Tables 2-13. Table 14 summarizes the Newman-Keuls results and shows the muscle rankings based on significance level for the two tasks.

The behavior of total power across time provided the only consistent predictor of task termination. In all cases, the amount of total power in each 16-second interval fell continuously for a discrete period of time prior to the time the subject could no longer move the controller out of detent against the spring's breakout forces (task termination). The number of seconds this phenomenon occurred across all subjects for each muscle is presented in Table 15, along with the standard deviations, means, ranges, and variances for each of the two tasks.

Table 1. F-Maximum Tests for Homogeneity of Variances

| | Left | Right |
|---|---|---|
| Biceps | 12.46 | 19.31 |
| Triceps | 10.05 | 4.74 |
| Brachioradialis | 130.87* | 11.99 |
| Flexor Carpi Ulnaris | 18.02 | 6.21 |
| Brachialis | 26.72** | 584.53* |
| Pronator Teres | 73.51* | 28.86** |

*$p<.01$
**$p<.05$

Table 2. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Left, *Biceps Brachii*

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 7991.1 | 47 | | | |
| SUBJECTS | 75.4 | 7 | | | |
| BANDS | 6767.0 | 1 | 6767.0 | 60.9 | # |
| TIMES | 2.3 | 2 | 1.1 | 0.8 | |
| BANDS x TIMES | 91.5 | 2 | 45.8 | 2.5 | |
| Error, Bands | 777.3 | 7 | 111.0 | | |
| Error, Times | 21.2 | 14 | 1.5 | | |
| Error, Bands x Times | 256.3 | 14 | 18.3 | | |

2a. Analysis of Variance

| | HIGH | LOW |
|---|---|---|
| | | 23.7* |

2b. Newman-Keuls Test for Difference Between Band Means

# = p<.001
* = p<.01

Table 3. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Right, Biceps Brachii

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 7403.5 | 47 | | | |
| SUBJECTS | 352.9 | 7 | | | |
| BANDS | 4174.6 | 1 | 4174.6 | 21.0 | ## |
| TIMES | 134.2 | 2 | 67.1 | 10.4 | ## |
| BANDS x TIMES | 720.5 | 2 | 360.3 | 9.4 | ## |
| Error, Bands | 1391.0 | 7 | 198.7 | | |
| Error, Times | 90.7 | 14 | 6.5 | | |
| Error, Bands x Times | 539.6 | 14 | 38.5 | | |

3a. Analysis of Variance

| Start | E>5 | End |
|---|---|---|
| | 1.2 | 4.0* |
| | | 2.8* |

3b. Newman-Keuls Test for Differences Among Time Means

| HIGH | LOW |
|---|---|
| | 18.6* |

3c. Newman-Keuls Test for Difference Between Band Means

## = p<.005
 * = p<.01

Table 4. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Left, <u>Triceps Brachii</u>

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 8047.7 | 47 | | | |
| SUBJECTS | 116.8 | 7 | | | |
| BANDS | 4812.3 | 1 | 4812.3 | 68.4 | # |
| TIMES | 175.5 | 2 | 87.7 | 12.7 | # |
| BANDS x TIMES | 1704.5 | 2 | 852.3 | 18.4 | # |
| Error, Bands | 492.7 | 7 | 70.4 | | |
| Error, Times | 96.4 | 14 | 6.9 | | |
| Error, Bands x Times | 649.4 | 14 | 46.4 | | |

4a. Analysis of Variance

| Start | E>5 | End |
|---|---|---|
| | 3.7* | 4.4* |
| | | 0.7 |

4b. Newman-Keuls Test for Differences Among Time Means

| HIGH | LOW |
|---|---|
| | 20.0* |

4c. Newman-Keuls Test for Difference Between Band Means

# = p<.001
* = p<.01

Table 5. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Right, *Triceps Brachii*

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 4631.1 | 47 | | | |
| SUBJECTS | 359.7 | 7 | | | |
| BANDS | 717.1 | 1 | 717.1 | 7.1 | ** |
| TIMES | 533.0 | 2 | 266.5 | 14.1 | # |
| BANDS x TIMES | 1520.4 | 2 | 760.2 | 20.0 | # |
| Error, Bands | 703.2 | 7 | 100.5 | | |
| Error, Times | 264.8 | 14 | 18.9 | | |
| Error, Bands x Times | 533.2 | 14 | 38.1 | | |

5a. Analysis of Variance

| Start | E>5 | End |
|---|---|---|
| | 3.7** | 8.2* |
| | | 4.5** |

5b. Newman-Keuls Test for Differences Among Time Means

| HIGH | LOW |
|---|---|
| | 7.7** |

5c. Newman-Keuls Test for Difference Between Band Means

# = p<.001
* = p<.01
** = p<.05

Table 6. Analysis of Variance Test of Percent Power, Roll Left, Brachioradialis

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 4594.2 | 47 | | | |
| SUBJECTS | 280.5 | 7 | | | |
| BANDS | 1553.4 | 1 | 1553.4 | 5.5 | |
| TIMES | 12.0 | 2 | 6.0 | 0.4 | |
| BANDS x TIMES | 257.2 | 2 | 128.6 | 5.6 | ** |
| Error, Bands | 1969.8 | 7 | 281.4 | | |
| Error, Times | 196.7 | 14 | 14.1 | | |
| Error, Bands x Times | 324.5 | 14 | 23.2 | | |

** = $p < .05$

Table 7. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Right, <u>Brachioradialis</u>

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 1353.2 | 47 | | | |
| SUBJECTS | 185.8 | 7 | | | |
| BANDS | 100.1 | 1 | 100.1 | 1.4 | |
| TIMES | 166.5 | 2 | 83.2 | 19.0 | # |
| BANDS x TIMES | 240.7 | 2 | 120.4 | 19.9 | # |
| Error, Bands | 514.4 | 7 | 73.5 | | |
| Error, Times | 61.2 | 14 | 4.4 | | |
| Error, Bands x Times | 84.6 | 14 | 6.0 | | |

7a. Analysis of Variance

| Start | E>5 | End |
|---|---|---|
| | 3.3* | 4.4* |
| | | 1.1 |

7b. Newman-Keuls Test for Differences Among Time Means

# = p<.001
* = p<.01

Table 8. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Left, <u>Flexor Carpi Ulnaris</u>

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 1756.5 | 47 | | | |
| SUBJECTS | 179.4 | 7 | | | |
| BANDS | 254.0 | 1 | 254.0 | 5.3 | |
| TIMES | 192.9 | 2 | 96.5 | 18.0 | # |
| BANDS x TIMES | 407.4 | 2 | 203.7 | 9.1 | ## |
| Error, Bands | 335.9 | 7 | 48.0 | | |
| Error, Times | 75.1 | 14 | 5.4 | | |
| Error, Bands x Times | 311.9 | 14 | 22.3 | | |

8a. Analysis of Variance

| Start | End | E>5 |
|---|---|---|
| | 3.2* | 4.8* |
| | | 1.6 |

8b. Newman-Keuls Test for Differences Among Time Means

# = p<.001
## = p<.005
* = p<.01

Table 9. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Right, <u>Flexor Carpi Ulnaris</u>

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 953.0 | 47 | | | |
| SUBJECTS | 224.5 | 7 | | | |
| BANDS | 23.0 | 1 | 23.0 | 1.0 | |
| TIMES | 209.5 | 2 | 104.8 | 21.5 | # |
| BANDS x TIMES | 195.2 | 2 | 97.6 | 17.7 | # |
| Error, Bands | 155.1 | 7 | 22.2 | | |
| Error, Times | 68.4 | 14 | 4.9 | | |
| Error, Bands x Times | 77.4 | 14 | 5.5 | | |

9a. Analysis of Variance

| Start | E>5 | End |
|---|---|---|
| | 1.6 | 5.0* |
| | | 3.4* |

9b. Newman-Keuls Test for Differences Among Time Means

# = p<.001
* = p<.01

Table 10. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Left, Brachialis

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 6385.3 | 47 | | | |
| SUBJECTS | 506.6 | 7 | | | |
| BANDS | 4310.2 | 1 | 4310.2 | 46.0 | # |
| TIMES | 4.6 | 2 | 2.3 | 0.3 | |
| BANDS x TIMES | 104.6 | 2 | 52.3 | 1.1 | |
| Error, Bands | 656.0 | 7 | 93.7 | | |
| Error, Times | 129.1 | 14 | 9.2 | | |
| Error, Bands x Times | 674.1 | 14 | 48.2 | | |

10a. Analysis of Variance

| | HIGH | LOW |
|---|---|---|
| HIGH | | 18.9* |

10b. Newman-Keuls Test for Difference Between Band Means

# = p<.001
* = p<.01

Table 11. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Right, Brachialis

| SOURCE | SS | df | MS | F | p |
| --- | --- | --- | --- | --- | --- |
| TOTAL | 9540.6 | 47 | ——— | | |
| SUBJECTS | 1165.8 | 7 | | | |
| BANDS | 4686.4 | 1 | 4686.4 | 11.4 | ** |
| TIMES | 4.7 | 2 | 2.3 | 1.5 | |
| BANDS x TIMES | 399.3 | 2 | 199.7 | 7.5 | * |
| Error, Bands | 2888.2 | 7 | 412.6 | | |
| Error, Times | 22.2 | 14 | 1.6 | | |
| Error, Bands x Times | 374.0 | 14 | 26.7 | | |

11a. Analysis of Variance

| | HIGH | LOW |
| --- | --- | --- |
| | | 19.7* |

11b. Newman-Keuls Test for Difference Between Band Means

* = p<.01
** = p<.05

Table 12. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Left, *Pronator Teres*

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 2770.0 | 47 | | | |
| SUBJECTS | 425.6 | 7 | | | |
| BANDS | 225.7 | 1 | 225.7 | 1.5 | |
| TIMES | 101.2 | 2 | 50.6 | 8.8 | ## |
| BANDS x TIMES | 497.5 | 2 | 248.7 | 9.2 | ## |
| Error, Bands | 1060.4 | 7 | 151.5 | | |
| Error, Times | 80.8 | 14 | 5.8 | | |
| Error, Bands x Times | 379.0 | 14 | 27.1 | | |

12a. Analysis of Variance

| Start | End | E>5 |
|---|---|---|
| | 3* | 3.1* |
| | | 0.1 |

12b. Newman-Keuls Test for Differences Among Time Means

## =p<.005
* =p<.01

Table 13. Analysis of Variance and Newman-Keuls Tests of Percent Power, Roll Right, Pronator Teres

| SOURCE | SS | df | MS | F | p |
|---|---|---|---|---|---|
| TOTAL | 1039.1 | 47 | | | |
| SUBJECTS | 113.7 | 7 | | | |
| BANDS | 206.7 | 1 | 206.7 | 4.0 | |
| TIMES | 69.6 | 2 | 34.8 | 4.8 | ** |
| BANDS x TIMES | 100.9 | 2 | 50.4 | 8.6 | ## |
| Error, Bands | 364.8 | 7 | 52.1 | | |
| Error, Times | 101.4 | 14 | 7.2 | | |
| Error, Bands x Times | 82.0 | 14 | 5.9 | | |

13a. Analysis of Variance

| Start | E>5 | End |
|---|---|---|
| | 2.2** | 2.8** |
| | | 0.6 |

13b. Newman-Keuls Test for Differences Among Time Means

## = p<.005
** = p<.05

Table 14. Muscle Rankings Based on Newman-Keuls Tests of Ordered Means

## Bands

### Roll Left

1. Biceps
2. Triceps
3. Brachialis

### Roll Right

1. Biceps
2. Brachialis
3. Triceps

## Times

### Roll Left

End - Start:

1. Triceps
2. Flexor Carpi Ulnaris
3. Pronator Teres

Error > 5 - Start:

1. Flexor Carpi Ulnaris
2. Triceps
3. Pronator Teres

### Roll Right

End - Start:

1. Triceps
2. Flexor Carpi Ulnaris
3. Brachioradialis
4. Biceps
5. Pronator Teres

Error > 5 - Start:

1. Brachioradialis
2. Triceps
3. Pronator Teres

End - Error > 5:

1. Flexor Carpi Ulnaris
2. Biceps
3. Triceps

Table 15. Time in Seconds for TP Decline Prior to Task Termination

Roll Left

Muscles

| | BB | T | BR | FCU | B | PT |
|---|---|---|---|---|---|---|
| Total | 375 | 439 | 423 | 375 | 439 | 407 |
| Mean | 46.9 | 54.9 | 52.9 | 46.9 | 54.9 | 50.9 |
| Variance | 364.7 | 677.8 | 873.8 | 447.0 | 348.7 | 640.1 |
| Range | 10-70 | 10-102 | 10-103 | 10-72 | 24-86 | 10-88 |
| Standard Deviation | 19.1 | 26.0 | 29.6 | 21.1 | 18.7 | 25.3 |

Roll Right

Muscles

| | BB | T | BR | FCU | B | PT |
|---|---|---|---|---|---|---|
| Total | 574 | 379 | 494 | 478 | 430 | 398 |
| Mean | 71.8 | 47.4 | 61.8 | 59.8 | 53.8 | 49.8 |
| Variance | 1147.6 | 326.3 | 392.2 | 697.4 | 753.4 | 1258.5 |
| Range | 32-122 | 26-71 | 34-90 | 26-106 | 16-106 | 8-106 |
| Standard Deviation | 33.9 | 18.1 | 19.8 | 26.4 | 27.5 | 35.5 |

# CHAPTER V

## DISCUSSION

This study explored various derived measures of the electromyographic activity of six muscles during an isometric tracking task utilizing a spacecraft sidearm controller. The specific goals of this study were met. The results strongly support previous EMG PSD studies of fatigue in terms of power shifts from high to low frequency bands during tasks of this nature. Because of its exploratory nature, the results associated with prediction of task termination should be viewed as suggestive rather than definitive of relationships.

Examination of Tables 2-13 reveals that only the triceps had significant effects for both factors and the interaction term in both the roll right and left tasks, and the Table 14 rankings suggest that the triceps is the most consistently involved of the six muscles in the two tasks. In fact, only the triceps showed significance in all possible tests for simple effects. Therefore, if only one muscle could be monitored for the two tasks, the triceps is the most appropriate.

Monitoring of the TP measure for continuous decrease after peaking in an isometric task suggests that prediction of task termination is possible. The data in Table 15 show the TP decline for the triceps and brachialis muscles to be the earliest predictor for the roll left task and the biceps the earliest for the roll right condition. However, if the standard deviation is considered, the plus-and-minus-one standard deviation range of prediction would be between 36.2 and 73.6 for the brachialis, which would yield at least two 16-second averaging intervals for prediction for approximately 68% of the population if the techniques of this study were employed.

For the roll right task, the _biceps_ is the earliest predictor, and the _biceps_, _brachioradialis_, and _flexor carpi ulnaris_ all yield at least two sixteen-second averaging intervals for the plus-and-minus-one standard deviation case, with the _brachioradialis_ providing the longer time due to its smaller standard deviation.

With respect to the analyses of variance, the F-Max test results shown in Table 1 indicate enough heterogeneity of variance in the _brachialis_ and _pronator teres_ muscles for both tasks and in the _brachioradialis_ for the roll left task to suggest a high probability of making a Type I error if these data are used to support hypotheses or make assumptions. This condition is possibly due, in part, to the fact that both the _brachialis_ and the _pronator teres_ were difficult to palpate, which could have resulted in some electrode placement variance.

For the remaining cases, the _biceps_ and _triceps_ showed significant effects for both tasks in the frequency bands factor. The bands mean square corresponds to a comparison between the means for low and high bands averaged over the three levels of the times factor. The significance of the bands mean square supports the conclusion that the percentage of power measured is a function of the frequency range in which it is measured.

The times mean square was significant for the _triceps_ and _flexor carpi ulnaris_ in both tasks and for the _biceps_ and _brachioradialis_ in the roll right tasks. This represents a comparison between the means for the three time levels averaged over the two frequency bands. Significance in these cases supports the conclusion that the amount of power is a function of the time in which it was measured for these muscle/task combinations.

Interaction effects for the relevant muscles were significant for the _triceps_ and _brachioradialis_ in both tasks and for the _brachioradialis_ in the roll right task. Significance in these cases supports the hypothesis that measurement of percent power in low or high bands is not independent and, therefore, is a function of the time at which the measurement was taken. Graphs of the interaction effects can be extracted from Figures 3-6, if desired, by combining the low and high frequency plots for a given task and muscle.

The design and analysis used in this study require significant effects in all three cases to support the contention of muscle involvement being demonstrated by increasing power in the low frequency band and decreasing power in the high frequency band. Clearly, the _triceps_ in both tasks and the _biceps_ in the roll right task fulfill these requirements. Chaffin (1969b) and Kwatny et al. (1970) demonstrated similar involvement in their studies.

The _biceps_ in the roll left task tended to have a significantly higher percent power in the lower frequency band for the duration of the task. While such a result is indicative of involvement, it is not useful for predictive purposes.

The _flexor carpi ulnaris_ in both tasks and the _brachioradialis_ in the roll right task tended to have significantly and increasingly higher percent power in the low band and relatively constant percent power in the high band. This result, too, is consistent with the previous findings of Chaffin (1969a) and provides usable information to support the EMG-fatigue relationship.

The _flexor carpi ulnaris_ was the only muscle apparently to peak in involvement at the point of error increasing beyond 5°/second and subsequently to begin a return to the state representing non-involvement as manifested by a decreasing percent power in the lower frequency band. This supports the findings of Johansson (1970) where a high frequency increase in percent power occurred during recovery. Of the remaining muscles showing significant effects, the rate of increase of percent power in the low frequency band declined after the error increased beyond 5°/second (excepting the _biceps_ and _triceps_ for the roll right task), indicating fatigue onset and declining ability of the muscle to contribute (Figures 3-6). This is consistent with the findings of Lippold et al. (1960) where a transfer of function to deeper muscles was found during a sustained isometric task. These findings support the postulation that such a phenomenon was occurring, and that the point was eventually reached when the number of remaining functional motor units in various muscles was insufficient to continue the task. An attempt to monitor a potentially early onset of this condition was made by observing the difference in percent power between the low and high bands over time (Figures A-6 and A-7). No predictive relationships were obtained, however, and further investigation of this derived measure was set aside for future consideration.

The general pattern of muscle involvement is also similar to the findings of Stevens et al. (1973). After examining seven upper arm muscles of 15 men and 12 women during supination and pronation exercise at tension levels up to maximum voluntary contraction, they found the triceps and brachioradialis contributory with the biceps most involved in supination and the brachialis most involved in pronation, although a specific programmed pattern of involvement could not be developed for a given individual.

Figures A-8 through A-11 graphically depict the amount of power and power distribution across frequencies in each 16-second contiguous interval. In the example illustrated, the total average power in an interval started at a relatively low level and increased gradually in each interval for approximately two minutes. Figure A-12 shows the power averaged for this period (120 to 232 seconds). For two intervals the amount of power jumped sharply in the lower frequency band to a level of about 2-1/2 times the previous average (see Figures A-10 and A-13) and subsequently began a decline that continued to task termination at 338 seconds (see Figures A-10, A-11, A-14, and A-15). The beginning of the two intervals shown in Figures A-10 and A-13 also coincides with the time (248 seconds) at which error began diverging.

The phenomenon of TP approximating zero prior to task termination, while the subject was still able to overcome the spring force to some extent, is readily seen in Figures A-11 and A-15 for the period of approximately 312-336 seconds.

In addition, the information in Figures A-12 through A-15 provides insight into the care required when utilizing percent power to the exclusion of amount of power. In Figure A-13, the shift toward the lower frequencies is easily observable as fatigue sets in (measured by tracking error beginning to diverge and exceed 5°/second). In Figure A-12, total average power for approximately two minutes was 13, with 35% located in the lower frequency band. At the point where error divergence began, total average power was almost 36 for 32 seconds with about 61 percent in the lower band. From this point to task termination, approximately 60 percent power resided in the lower band, but after the next 48 seconds, total average power dropped to 9 and finally to about 0.1 for the final 26 seconds.

While the information is useful and these events are reliable in that they occurred in all subjects and all muscles, monitoring percent power only could lead to potentially erroneous conclusions of significant muscle involvement during the final moments of an isometric task carried to exhaustion (in this case, the final half minute).

These results establish that the bioelectric frontier continues to reveal useful information for the investigation of muscle involvement in isometric tasks. However, many factors contribute to variance in typical dependent variables in a study such as this. Whether an experiment is static or dynamic is important: If dynamic, the individual muscle exercise/rest duty cycle is significant; if static, the isotonic or isometric actions involved across muscles must be considered, while individual strength, percent MVC, training, motivation, and structural considerations such as biomechanical angles, restraint systems, and control resolution must be integrated into the experiment. Finally, the experimental protocol must be rigidly implemented and reported to minimize variance in comparisons across studies.

In general, the agreement with previous efforts and the new data suggested for physiological predictive purposes reinforce the potential utility of spectral analysis and electromyography in the human factors design areas of habitability, work space and tool design, in medical applications for measurement of pathologies and rehabilitation progress, in physical training for comparative studies of alternative methods, and in a constrained environment such as a pressurized space or diving suit to monitor the status and functional condition of muscular activity.

For EMG and spectral analysis to become a viable tool, however, much applied research must be accomplished to further define those measures specific to a given task to better automate the collection, reduction, and interpretation of data.

CHAPTER VI

# SUMMARY AND CONCLUSIONS

Electromyographic recordings of activity in six muscles for eight subjects were reduced with the Fast Fourier Transform to acquire power spectral density measurements in isometric tracking tasks. A prototype three-axis spacecraft handcontroller, spring-loaded to a center detent, was utilized in roll right and left tasks which were maintained to a point where the subjects could no longer move the controller out of detent against the spring force. Data were taken for the *biceps brachii*, *triceps brachii*, *brachioradialis*, *flexor carpi ulnaris*, *brachialis*, and *pronator teres* muscles.

The data were checked for general conformity to previous and potentially related EMG studies of fatigue, to determine which single muscle might best serve as an overall indicator of involvement in both tasks, and to determine if tracking performance degradation could be predicted.

The goals of the study were met: the results showed general agreement with previous studies; a single muscle common to both tasks was identified; and a measure potentially useful for predicting performance decrement was identified.

Specific to the results of this study, several conclusions can be drawn:

1) The tasks utilized were suitable for identifying muscle involvement. A complete analysis of participation in the tasks for given muscles would be difficult if the task were terminated prior to the point of exhaustion.

2) Sufficient homogeneity existed to perform analyses of variance on the biceps, triceps, and flexor carpi ulnaris for the roll left task, and on the biceps, triceps, brachioradialis, and flexor carpi ulnaris for the roll right task. In the remaining cases--i.e., the brachialis and pronator teres for both tasks, and the brachioradialis for the roll left task--there is a high degree of probability of attributing significance where, in fact, none exists.
3) Of the six muscles, only the triceps was significantly involved in all the possible tests for simple effects across the two tasks.
4) Of the various derived statistics investigated, only total power was a consistent predictor of task termination.
5) The triceps and brachialis muscles for roll left and the biceps for the roll right task were the earliest predictors of task termination. However, for the population studied, the variances for the brachialis for roll left, and the biceps, brachioradialis, and flexor carpi ulnaris for roll right were small enough to permit prediction as early as two averaging intervals prior to termination for a plus-and-minus-one standard deviation range.
6) Percent power generally remains relatively constant after fatigue has set in, even though total power has dropped to a point where the muscle is no longer contributing to the task. Therefore, total power and percent power in a given frequency band should be considered together even where adjustments for calibration differences across recording channels and subjects have not been accomplished.

7) EMG recording and PSD analysis are useful tools in the investigation of muscular behavior in work tasks and have potentially wide applications across disciplines.

Further applied research is required to identify measures sensitive to specific needs of the investigator. For example, the prediction of tracking error divergence would be useful knowledge in various work tasks, or the point when function transfers to deeper muscles would be useful information in hardware design. Furthermore, much additional work is needed to automate the data collection/reduction process to provide nearer real-time output to the user before maximum application of the potential can be realized.

# BIBLIOGRAPHY

# BIBLIOGRAPHY


Bartley, S. H. Fatigue: Mechanism and management. Springfield, Illinois: Charles Thomas, 1965.

Basmajian, J. V. Muscles alive, their functions revealed by electromyography, second edition. Baltimore: Williams and Wilkins Co., 1967.

Bassett, D. L. A Stereoscopic atlas of human anatomy. Portland, Oregon: Sawyers, Inc., 1960.

Bouisset, S., and Maton, B. Quantitative relationship between surface EMG and intramuscular electromyographic activity in voluntary movement. American Journal of Physical Medicine, 1972, 51(6):285-295.

Broman, H., Magnusson, R., Petersen, I., and Ortengren, R. Vocational electromyography. New Developments in Electromyography and Clinical Neurophysiology, 1973, 1:656-664.

Bruning, J. L. and Kintz, B. L. Computational handbook of statistics. Glenview: Scott, Foresman & Co., 1968.

Cenkovich, F. S., and Gersten, J. W. Fourier analysis of the normal human electromyogram. American Journal of Physical Medicine, 1963, 42:192-204.

Chaffin, D. Surface electromyography frequency analysis as a diagnostic tool. Journal of Occupational Medicine, 1969a, 11(3):109-115.

Chaffin, D. Physical fatigue: What it is - how it is predicted. Journal of Methods - Time Measurement, 1969b, 14(3):20-28.

Chaffin, D. Electromyography - a method of measuring local muscle fatigue. Journal of Methods - Time Measurement, 1969c, 14(2):29-36.

Chaffin, D. Localized muscle fatigue - defintion and measurement. Journal of Occupational Medicine, 1973, 15(4):346-354.

Chapman, A. E. and Troup, J. D. G. The effect of increased maximal strength on the integrated electrical activity of lumbar erectores spinae. Electromyography, 1969, 9:263-280.

Currier, D. P. Measurement of muscle fatigue. Physical Therapy 1970, 49(7):724-730.

deVries, H. A. Efficiency of electrical activity as a physiological measure of the functional state of muscle tissues. American Journal of Physical Medicine, 1968a, 47(1):10-22.

deVries, H. A. Method for evaluation of muscle fatigue and endurance from electromyographic fatigue curves. American Journal of Physical Medicine, 1968b, 47(3):125-135.

Eason, R. G. Electromyographic study of local and generalized muscle impairment. Journal of Applied Physiology, 1960, 15(3):479-482.

Edwards, A. L. Experimental design in psychological research, fourth edition. New York: Holt, Rinehart and Winston, Inc., 1972.

Edwards, R. G., and Lippold, O. J. C. The relation between force and integrated electrical activity in fatigued muscle. Journal of Physiology, 1956, 132:677-681.

Geddes, L. A. Electrodes and the measurement of bioelectric events. New York: John Wiley & Sons, Inc., 1972.

Geddes, L. A., and Baker, L. E. Principles of applied biomechanical instrumentation. New York: John Wiley and Sons, Inc., 1968.

Hartley, H. O. The maximum F-ratio as a shortcut test for heterogeneity of variance. Biometrika, 1950, 37:308-312.

Hinson, M. M. Kinesiology. Dubuque, Iowa, W. C. Brown, Co., 1977.

Johansson, S., Larsson, L. E., and Ortengren, R. An automated method for the frequency analysis of myoelectric signals evaluated by an investigation of the spectral changes following strong sustained contractions. Medical and Biological Engineering, 1970, 8:257-264.

Kadefors, R., Petersen, I., and Herberts, P. Muscular reaction to welding work: An electromyographic investigation. Ergonomics, 1976, 19(5):543-558.

Kendall, H. O., Kendall, F. P., and Wadsworth, G. E. Muscles: Testing and function. Baltimore: Williams and Wilkins Co., 1971.

Komi, P. V., and Viitasalo, J. H. T. Signal characteristics of EMG at different levels of muscle tension. Acta Physiologica Scandinavica, 1976, 96:267-276.

Kuroda, E., Klissouras, V., and Milsum, J. H. Electrical and metabolic activities and fatigue in human isometric contraction. Journal of Applied Physiology, 1970, 29(3):358-367.

Kwatny, E., Thomas, D. H., and Kwatny, H. G. An application of signal processing techniques to the study of myoelectric signals. IEEE Transactions on Bio-Medical Engineering, 1970, 17(4):303-312.

LaFevers, E. V. Power spectral density analysis of the electromyogram from a work task performed in a full pressure suit. Doctoral Dissertation, University of Houston, Houston, Texas, 1974. (Also published as NASA Technical Memorandum TMX-58136, National Technical Information Services, Springfield, Va., 1974a).

LaFevers, E. V. Evaluation of localized muscle fatigue using power spectral density analysis of the electromyogram. Proceedings, 18th Annual Human Factors Society Meeting, October 1974b.

LaFevers, E. V., Nicogossian, A. E., Hoffler, G. W., Hursta, W., and Baker, J. Spectral analysis of skeletal muscle changes resulting from 59 days of weightlessness in Skylab II. NASA Technical Memorandum TMX-58171, National Technical Information Service, Springfield, Va., 1975.

Lance, B. M., and Chaffin, D. B. The effect of prior muscle exertions on simple movements. Human Factors, 1971, 13(4):355-361.

Lippold, O. C. J. The relation between integrated action potentials in a human muscle and its isometric tension. Journal of Physiology, 1952, 117:492-499.

Lippold, O. C. J., Redfern, J. W. T., and Vuco, J. The electromyography of fatigue. Ergonomics, 1952, 3(2):121-131.

Lloyd, A. J. Surface electromyography during sustained exertions. Journal of Applied Physiology, 1971, 30(5):713-719.

Moore, J. W. An electromyographic investigation of localized arm muscle fatigue in humans. Doctoral Dissertation, University of Houston, Houston, Texas, 1977.

Ortengren, R., Andersson, G., Broman, H., Magnusson, R., and Petersen, I. Vocational electromyography: Studies of localized muscle fatigue at the assembly line. Ergonomics, 1975, 18(2):157-174.

Ortengren, R. A filter bank analyzer with simultaneous readout for the evaluation of dynamic myoelectric signal power spectra. Medical and Biological Engineering, 1975, 13:561-569.

Schanne, F. J., and Chaffin, D. B. The effects of skin resistance and capacitance coupling on EMG amplitude and power spectra. Electromyography, 1970, 3:273-286.

Scherrer, J., and Bourguinon, A. Changes in the electromyogram produced by fatigue in man. American Journal of Physical Medicine, 1959, 38:170-180.

Stevens, A., Stijns, H., Reybrouck, T., Bonte, G., Michels, T., Rousselle, N., Roelandts, P., Krauss, E., and Verheyen, G. A polymyographical study of the arm muscles at gradual isometric loadings. Electromyographic and Clinical Neurophysiology, 1973, 13:465-476.

Viitasalo, J. H. T., and Komi, P. V. Signal characteristics of EMG with special reference to reproducibility of measurements. Acta Physiologica Scandinavica, 1975, 93:531-539.

Wells, K. F., and Luttgens, K. Kinesiology. Philadelphia: W. B. Saunders Co., 1976.

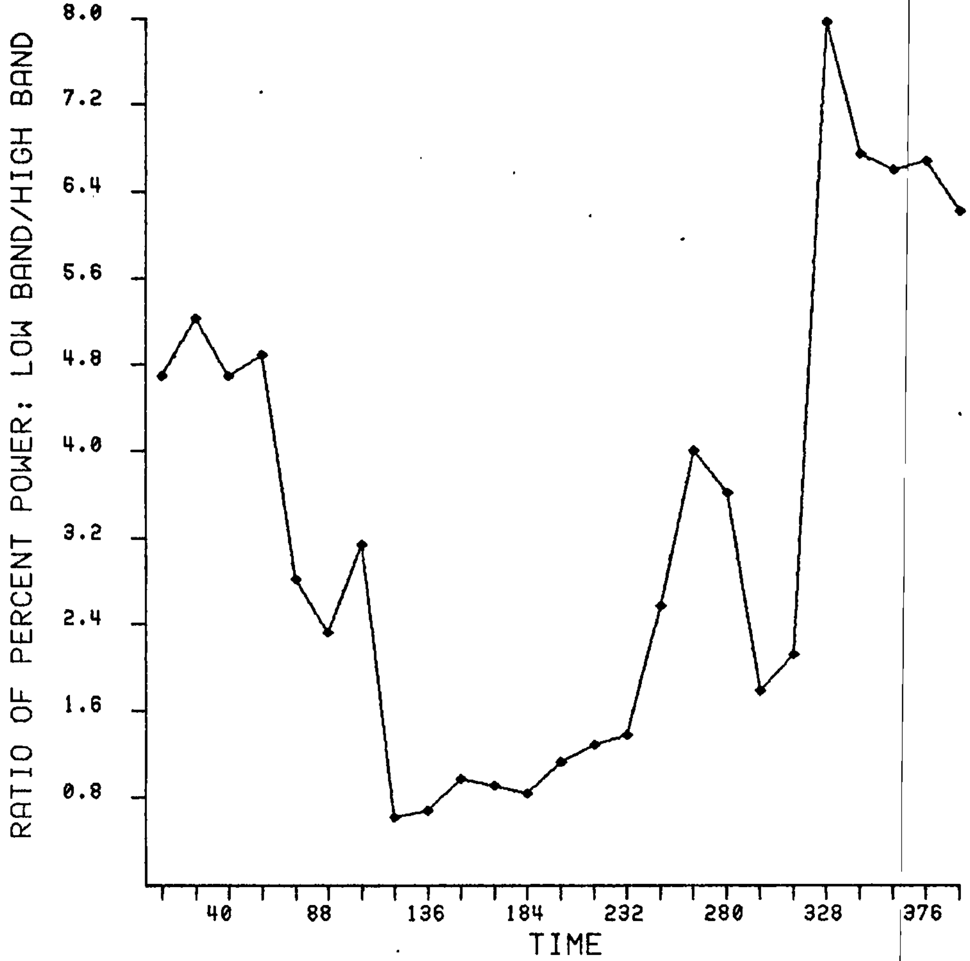
Winer, B. J. Statistical principles in experimental design. New York: McGraw Hill, 1962.

Zuniga, E. N., and Simons, D. G. Nonlinear relationship between averaged electromyogram potential and muscle tension in normal subjects. Archives of Physical Medicine and Rehabilitation, 1969, 50:613-620.

# APPENDIX A

# EMG DATA

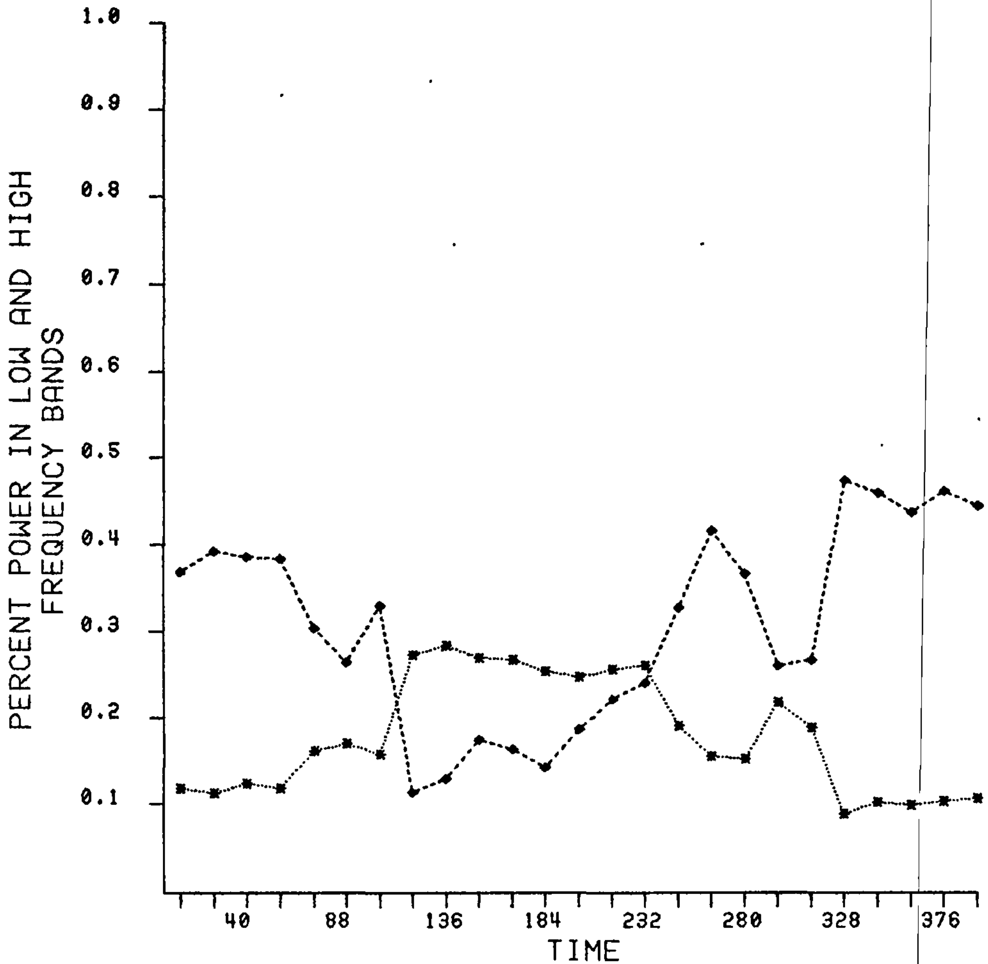
# SAMPLE PLOT FORMATS




SPECTRAL INDEX

◆ TRICEPS

SUBJ NO. 1
ROLL RIGHT
FIGURE A-1

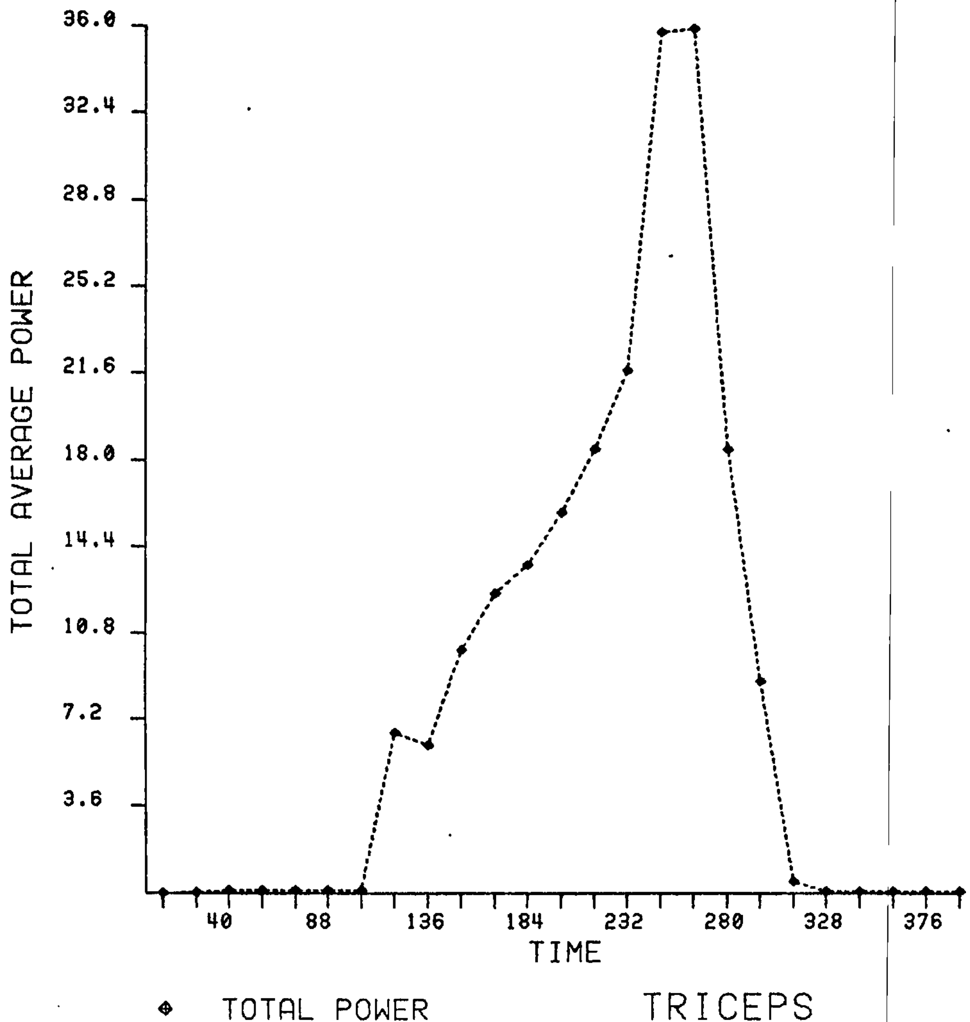



◆ X(I) 12-50HZ
✱ X(I) 62-100 HZ

TRICEPS ROLL RIGHT

SUBJ NO. 1

FIGURE A-2

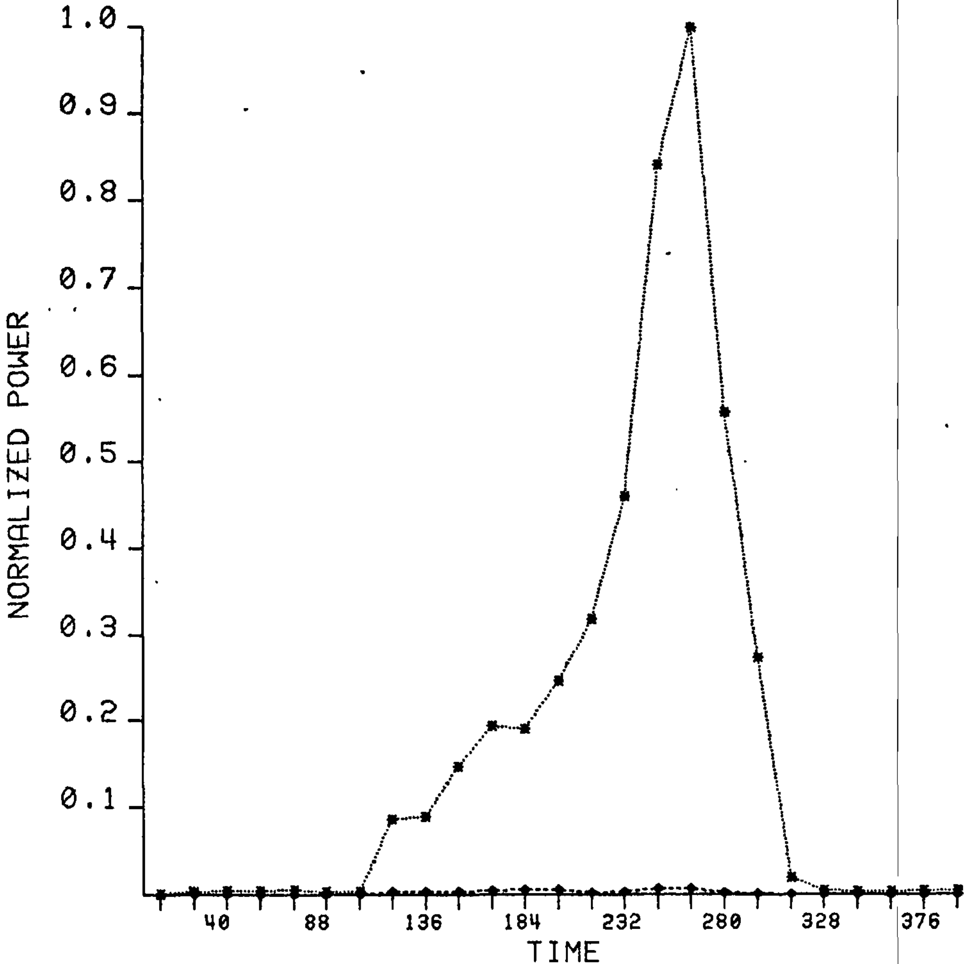



TRICEPS

SUBJ NO. 1

FIGURE A-3

ROLL RIGHT




◆ MINIMUM POWER

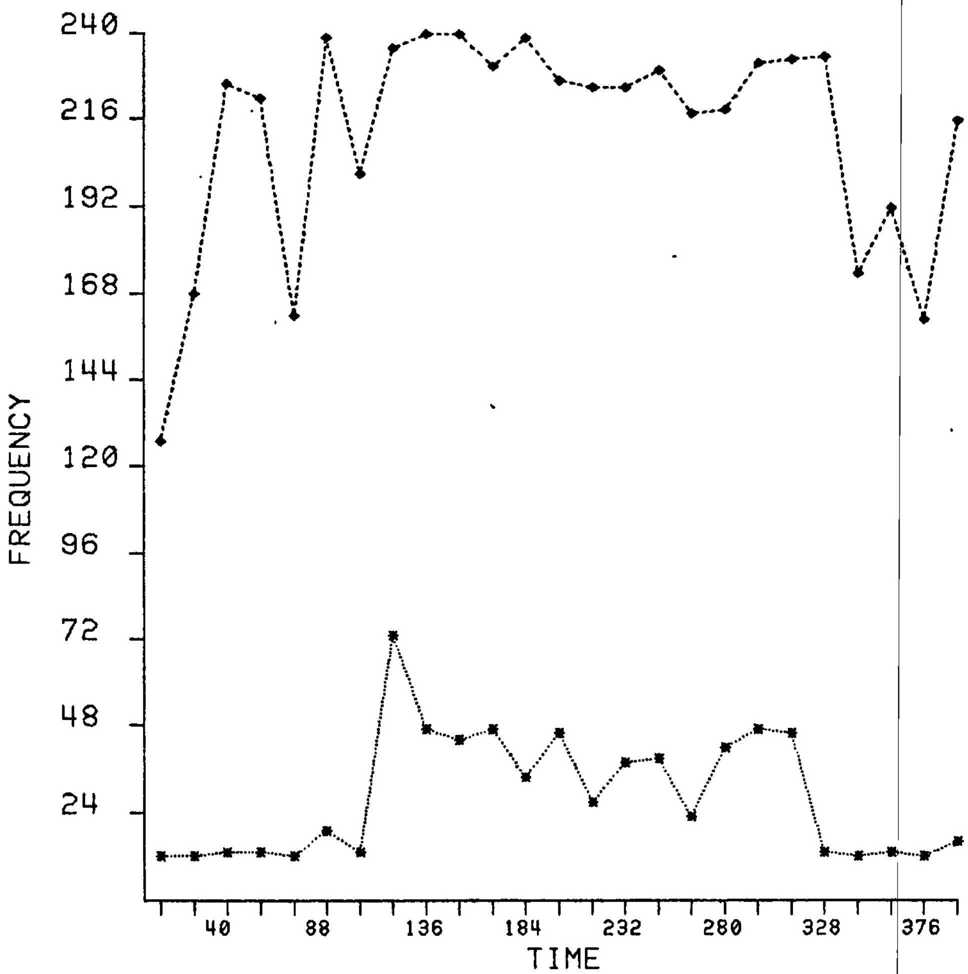
* MAXIMUM POWER

TRICEPS

SUBJ NO. 1

FIGURE A-4

ROLL RIGHT



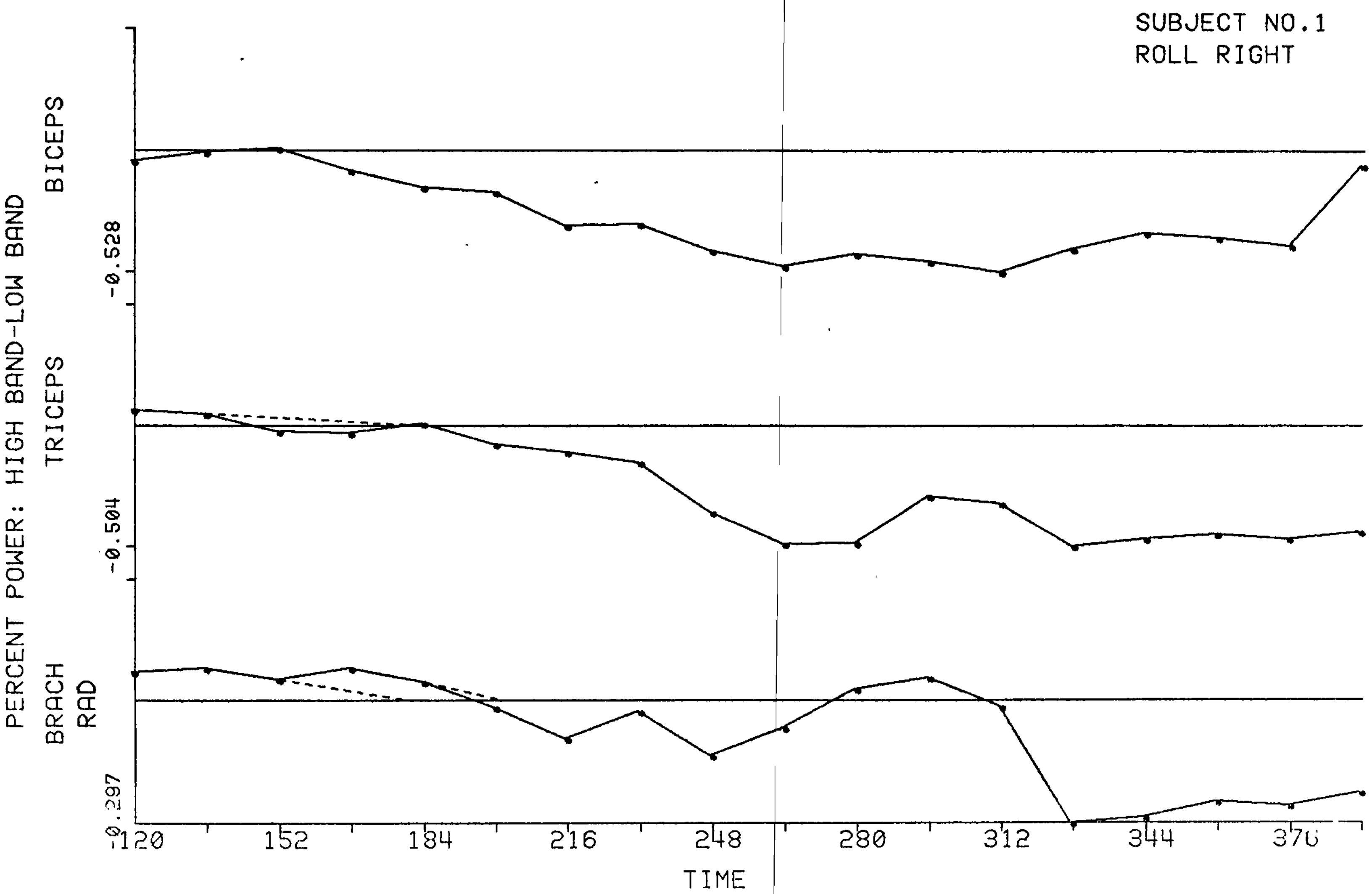

◆ FREQ(MIN.POWER)
＊ FREQ(MAX.POWER)

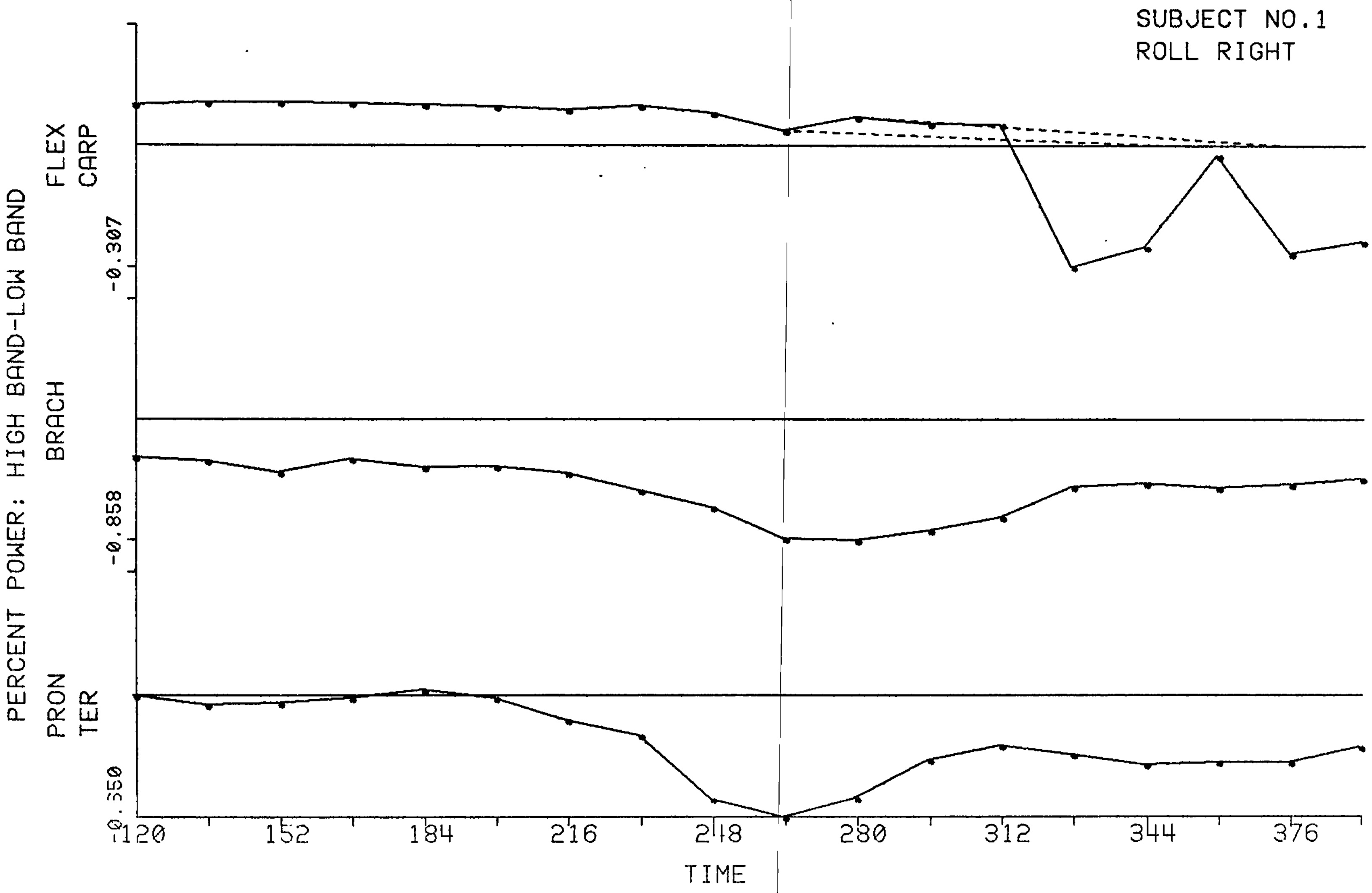
TRICEPS

ROLL RIGHT

SUBJ NO. 1

FIGURE A-5


SUBJECT NO.1
ROLL RIGHT
PERCENT POWER: HIGH BAND-LOW BAND
BICEPS
-0.528
TRICEPS
-0.504
BRACH
RAD
0.297
120
152
184
216
248
280
312
344
376
TIME
FIGURE A-6



SUBJECT NO.1
ROLL RIGHT
PERCENT POWER: HIGH BAND-LOW BAND
FLEX CARP
-0.307
BRACH
-0.858
PRON TER
0.550
120
152
184
216
248
280
312
344
376
TIME


FIGURE A-7

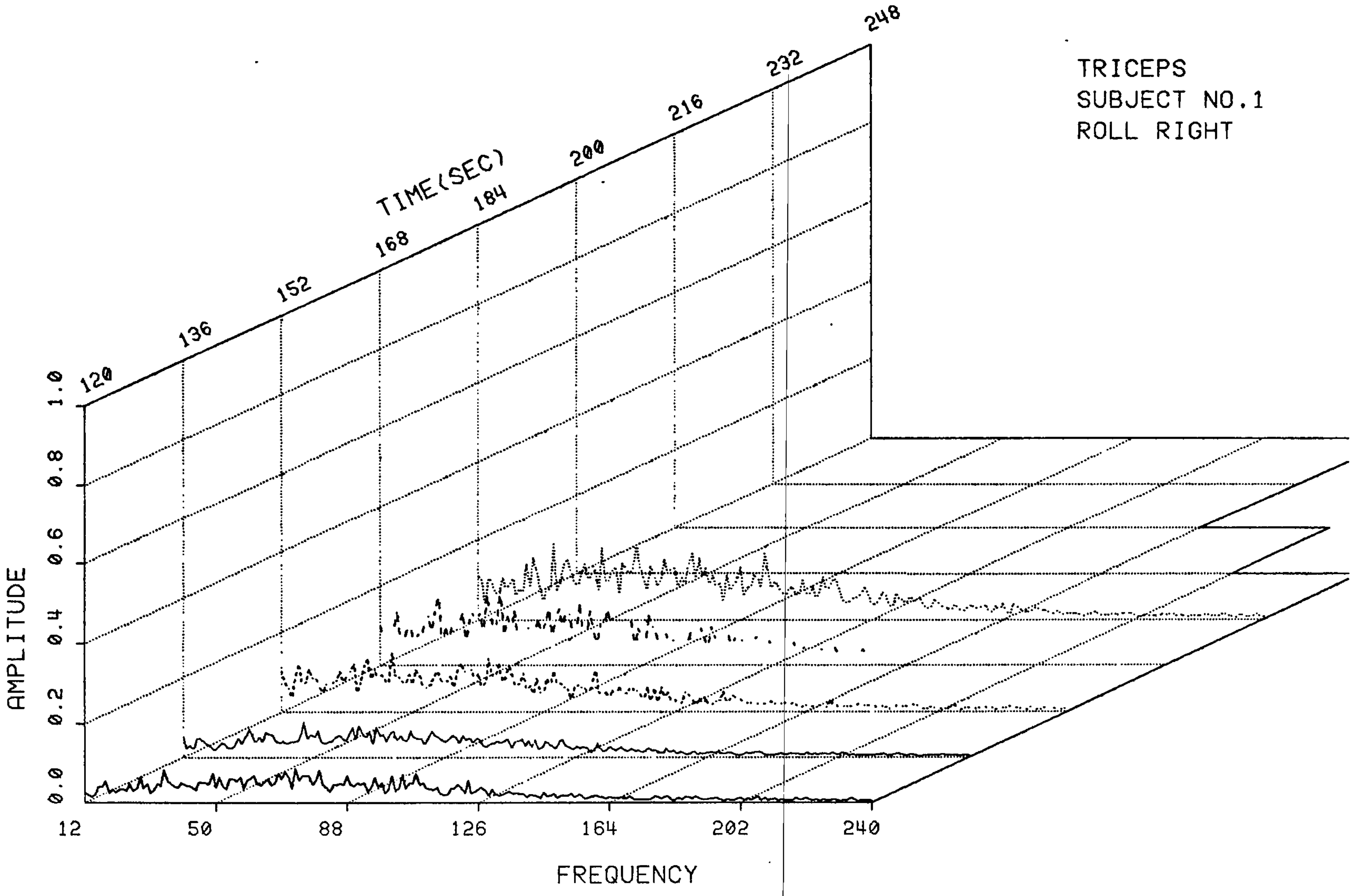

TRICEPS
SUBJECT NO.1
ROLL RIGHT
TIME(SEC)
120
136
152
168
184
200
216
232
248
AMPLITUDE
0.0
0.2
0.4
0.6
0.8
1.0
12
50
88
126
164
202
240
FREQUENCY

FIGURE A-8

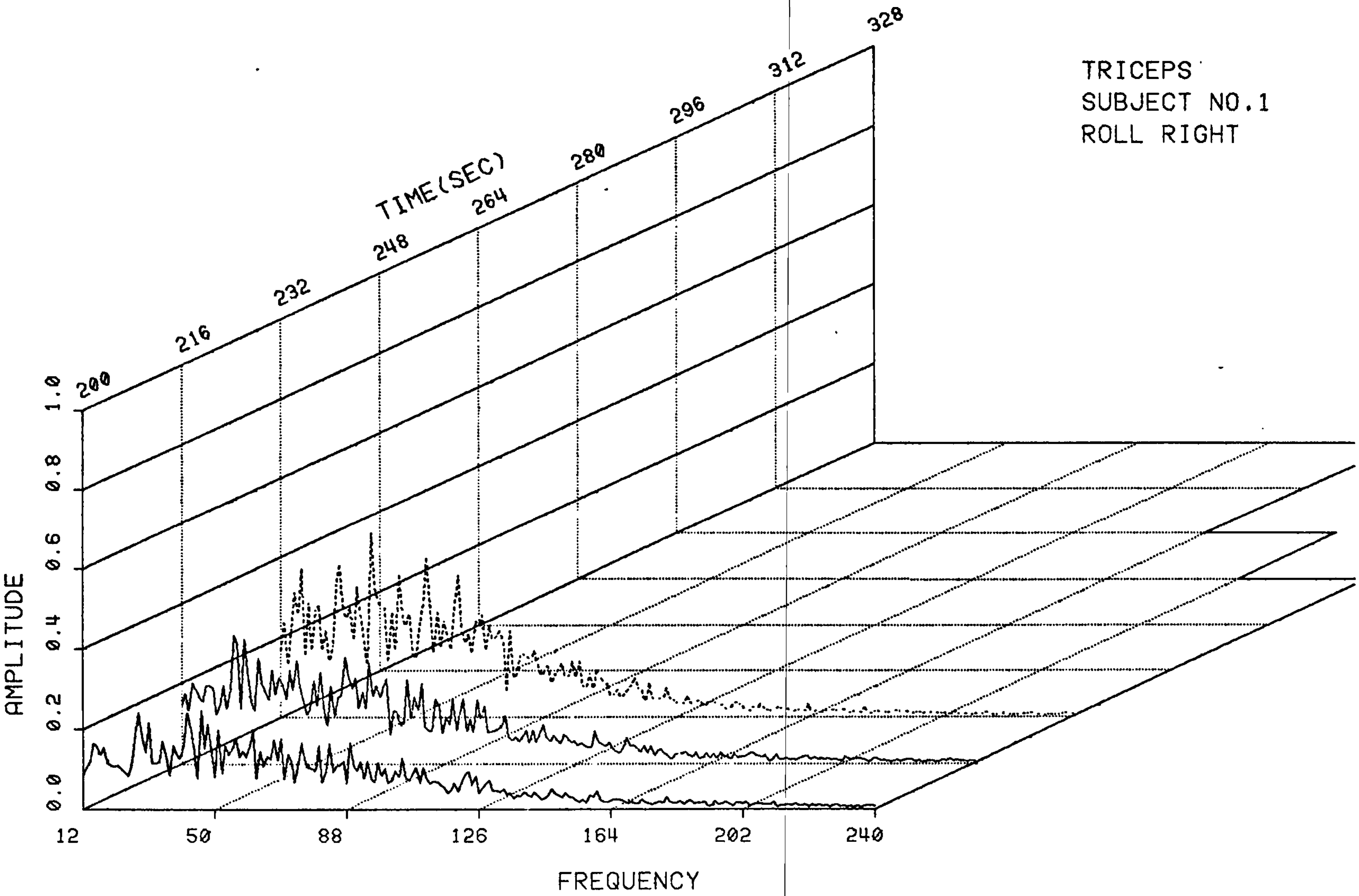
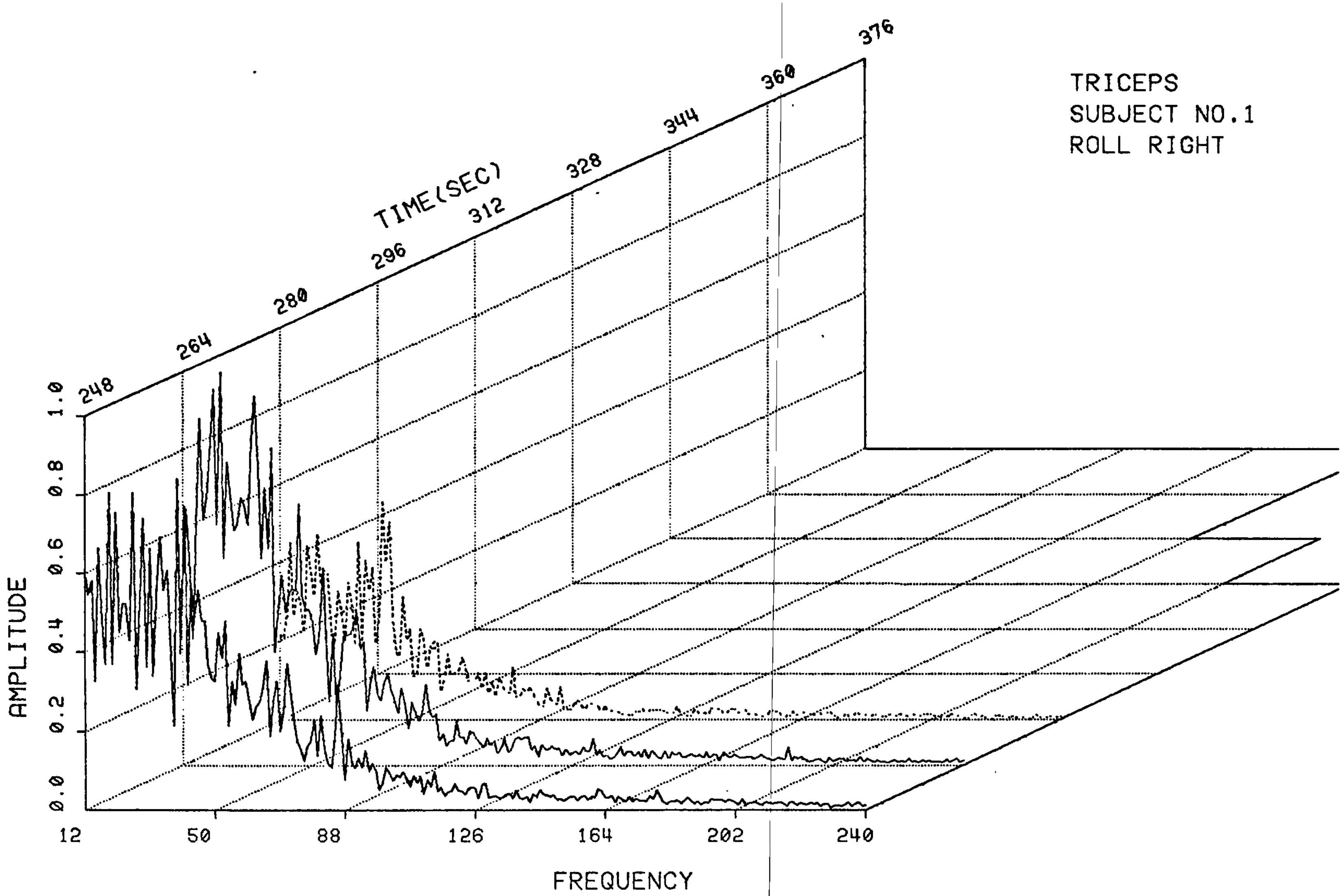

TRICEPS
SUBJECT NO.1
ROLL RIGHT
TIME(SEC)
200
216
232
248
264
280
296
312
328
AMPLITUDE
0.0
0.2
0.4
0.6
0.8
1.0
12
50
88
126
164
202
240
FREQUENCY

FIGURE A-9


TRICEPS
SUBJECT NO.1
ROLL RIGHT
TIME(SEC)
248
264
280
296
312
328
344
360
376
AMPLITUDE
0.0
0.2
0.4
0.6
0.8
1.0
12
50
88
126
164
202
240
FREQUENCY
FIGURE A-10

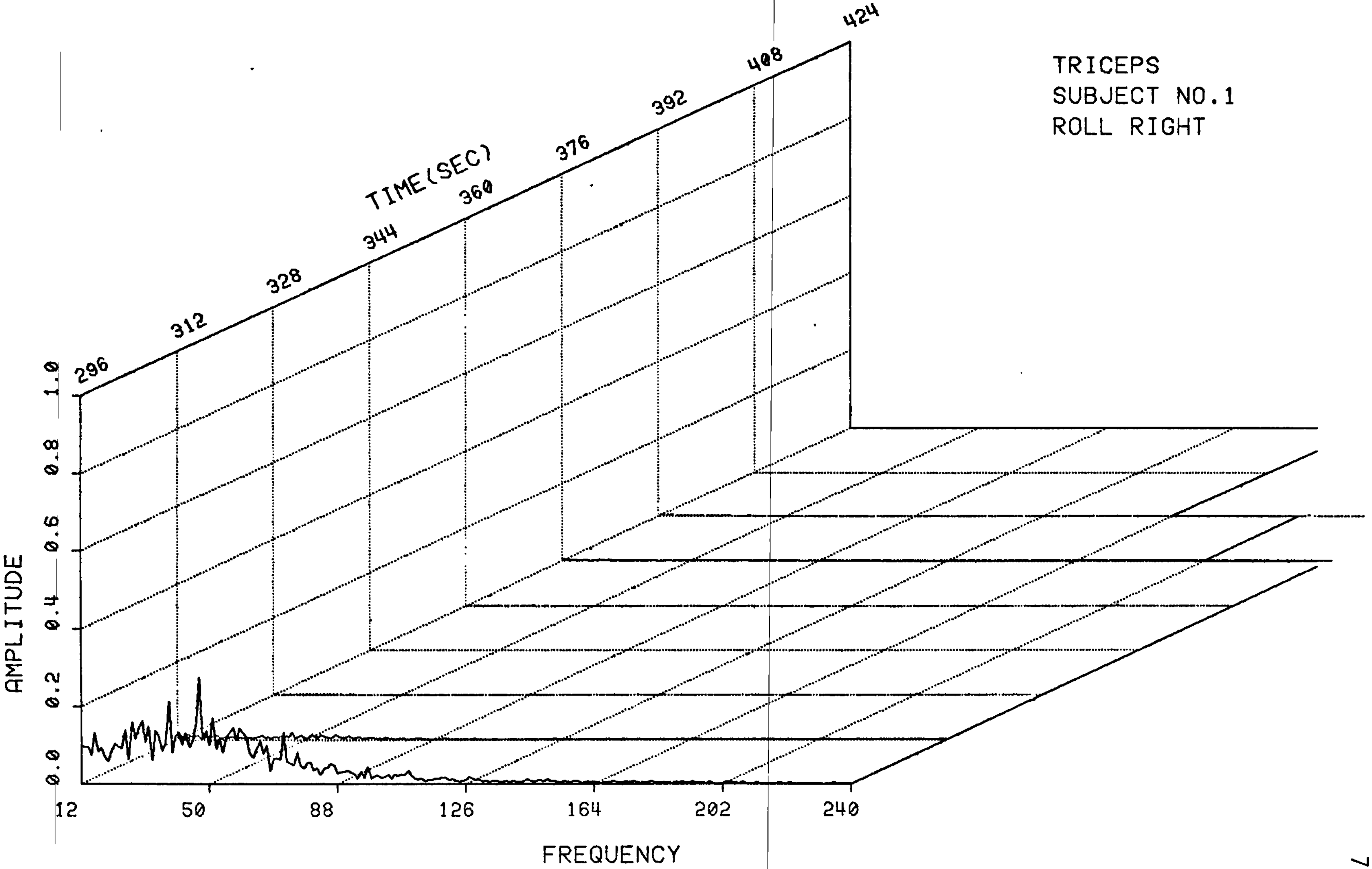

TRICEPS
SUBJECT NO.1
ROLL RIGHT
TIME(SEC)
296
312
328
344
360
376
392
408
424
AMPLITUDE
0.0
0.2
0.4
0.6
0.8
1.0
12
50
88
126
164
202
240
FREQUENCY

FIGURE A-11

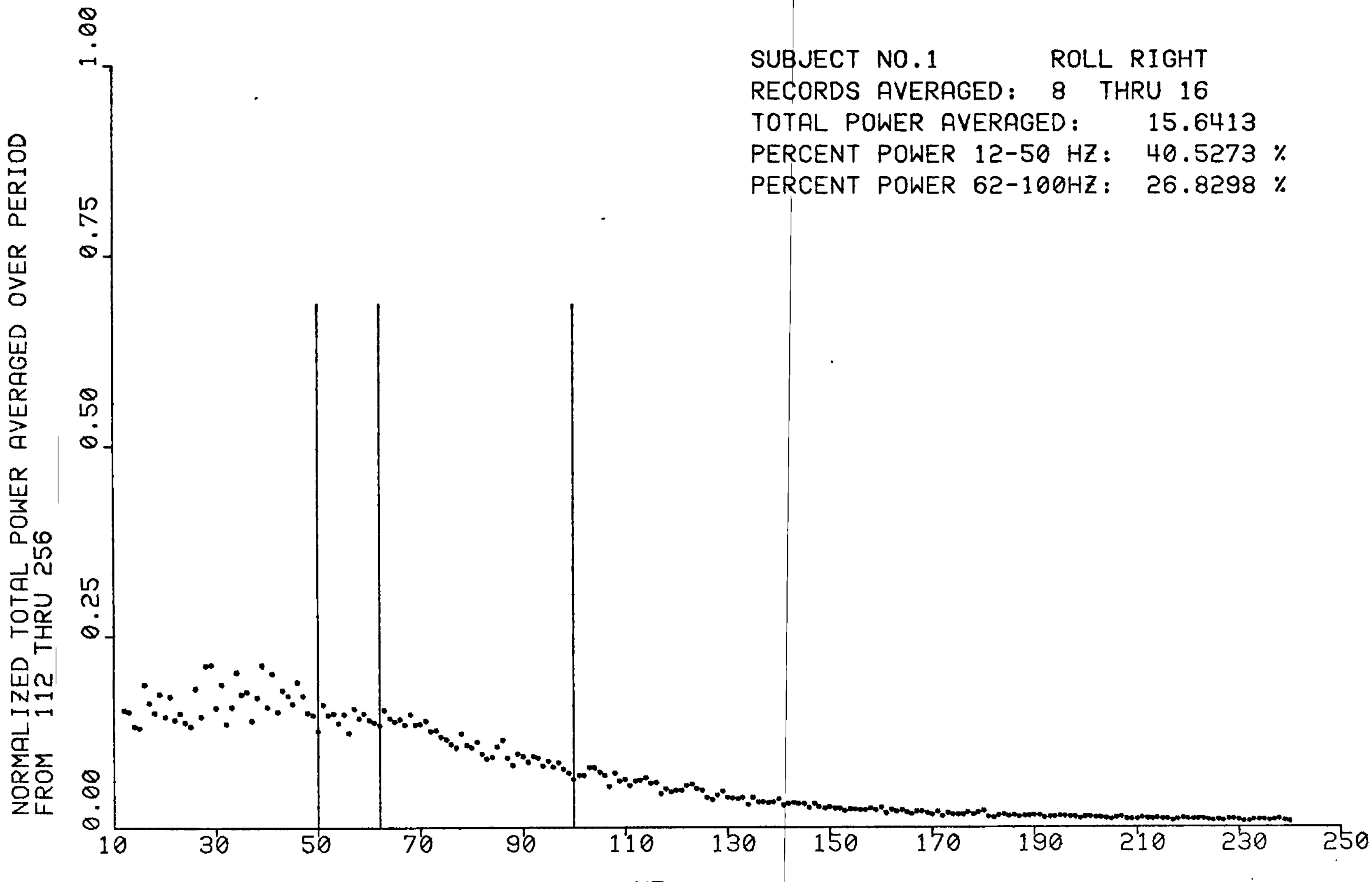

SUBJECT NO.1 ROLL RIGHT
RECORDS AVERAGED: 8 THRU 16
TOTAL POWER AVERAGED: 15.6413
PERCENT POWER 12-50 HZ: 40.5273 %
PERCENT POWER 62-100HZ: 26.8298 %
NORMALIZED TOTAL POWER AVERAGED OVER PERIOD
FROM 112 THRU 256
0.00
0.25
0.50
0.75
1.00
10
30
50
70
90
110
130
150
170
190
210
230
250
HZ

FIGURE A-12

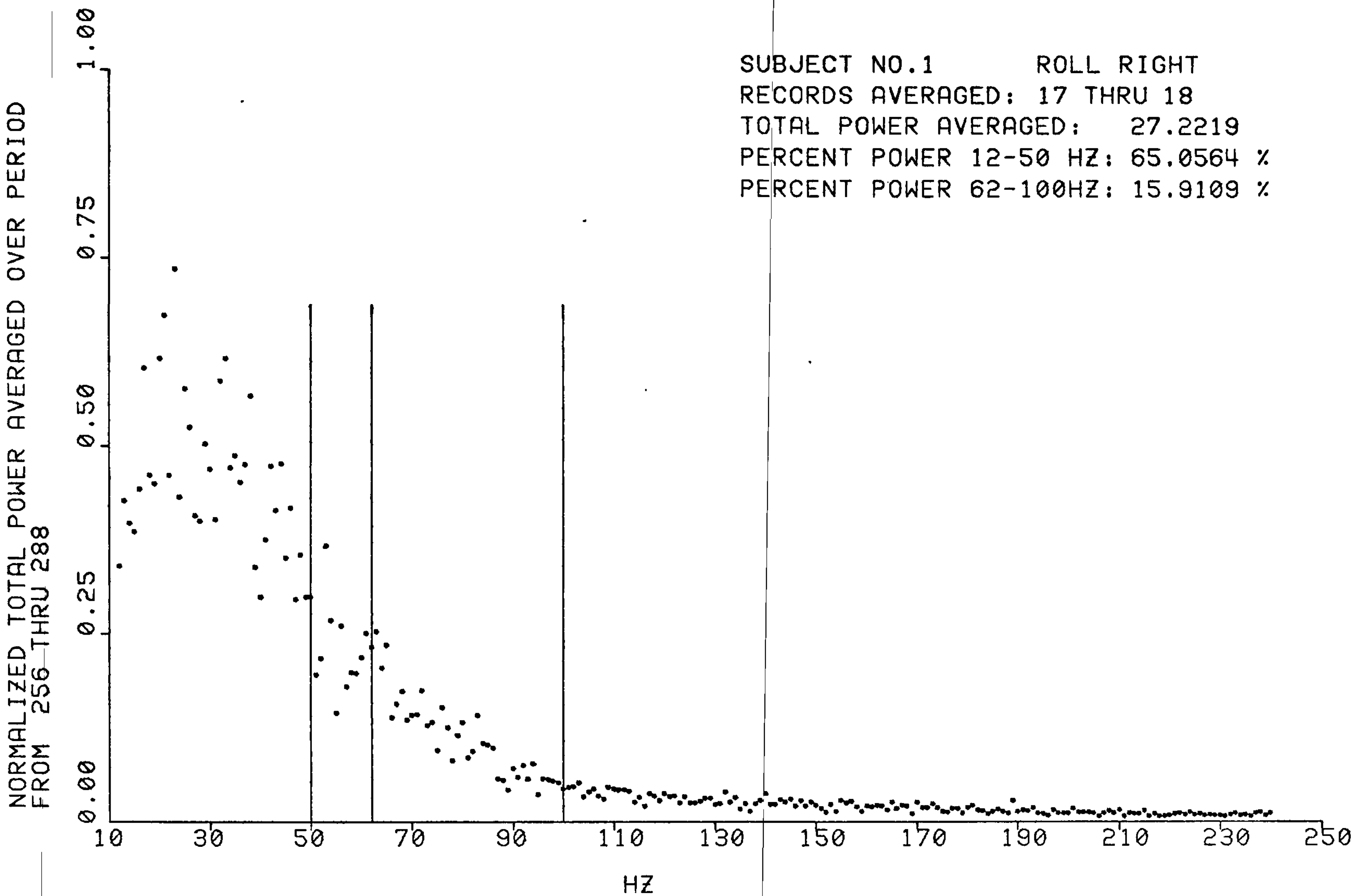

SUBJECT NO.1 ROLL RIGHT
RECORDS AVERAGED: 17 THRU 18
TOTAL POWER AVERAGED: 27.2219
PERCENT POWER 12-50 HZ: 65.0564 %
PERCENT POWER 62-100HZ: 15.9109 %
NORMALIZED TOTAL POWER AVERAGED OVER PERIOD
FROM 256 THRU 288
0.00
0.25
0.50
0.75
1.00
10
30
50
70
90
110
130
150
170
190
210
230
250
HZ

FIGURE A-13

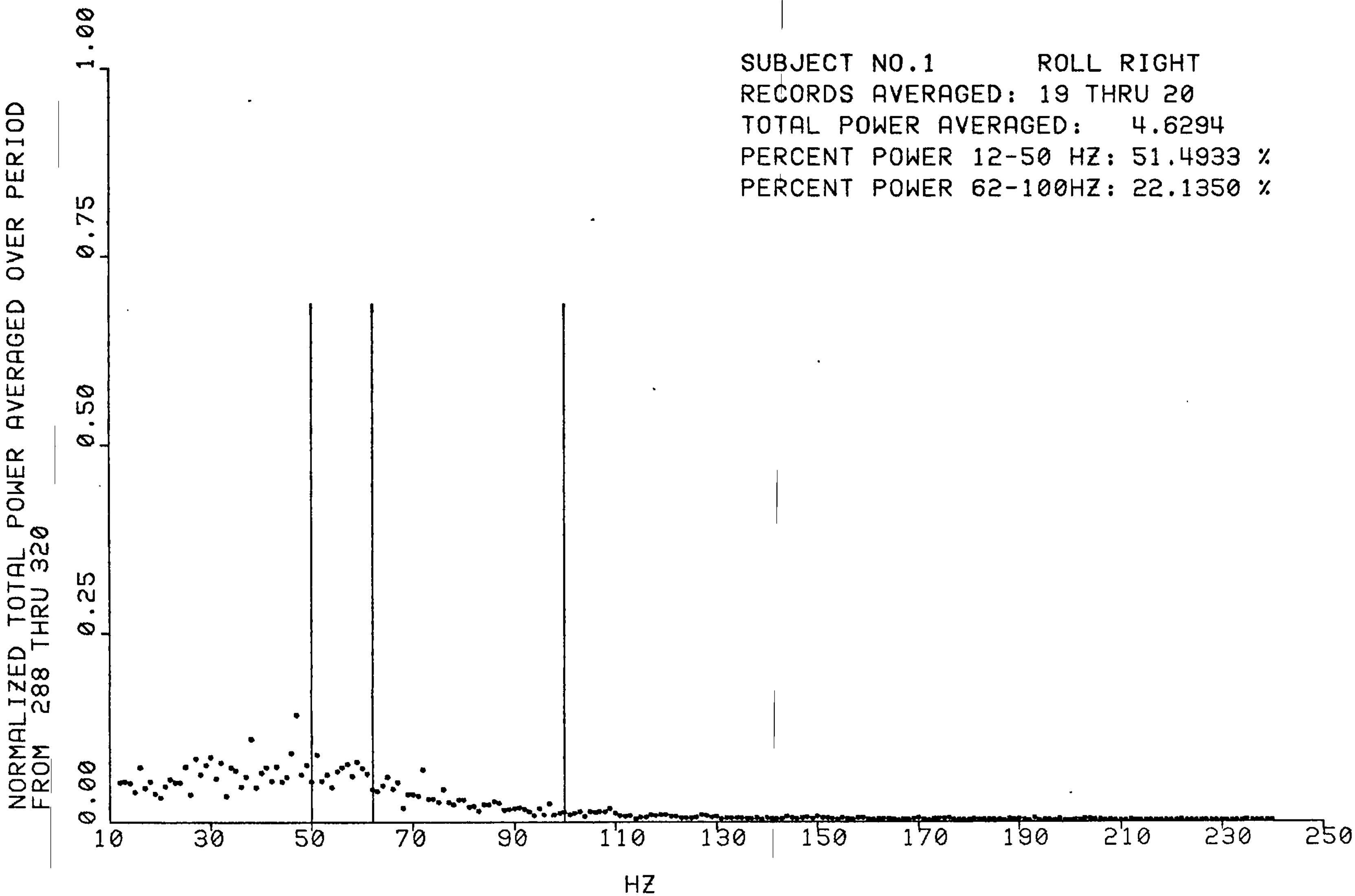

SUBJECT NO.1 ROLL RIGHT
RECORDS AVERAGED: 19 THRU 20
TOTAL POWER AVERAGED: 4.6294
PERCENT POWER 12-50 HZ: 51.4933 %
PERCENT POWER 62-100HZ: 22.1350 %
NORMALIZED TOTAL POWER AVERAGED OVER PERIOD
FROM 288 THRU 320
0.00
0.25
0.50
0.75
1.00
10
30
50
70
90
110
130
150
170
190
210
230
250
HZ

FIGURE A-14

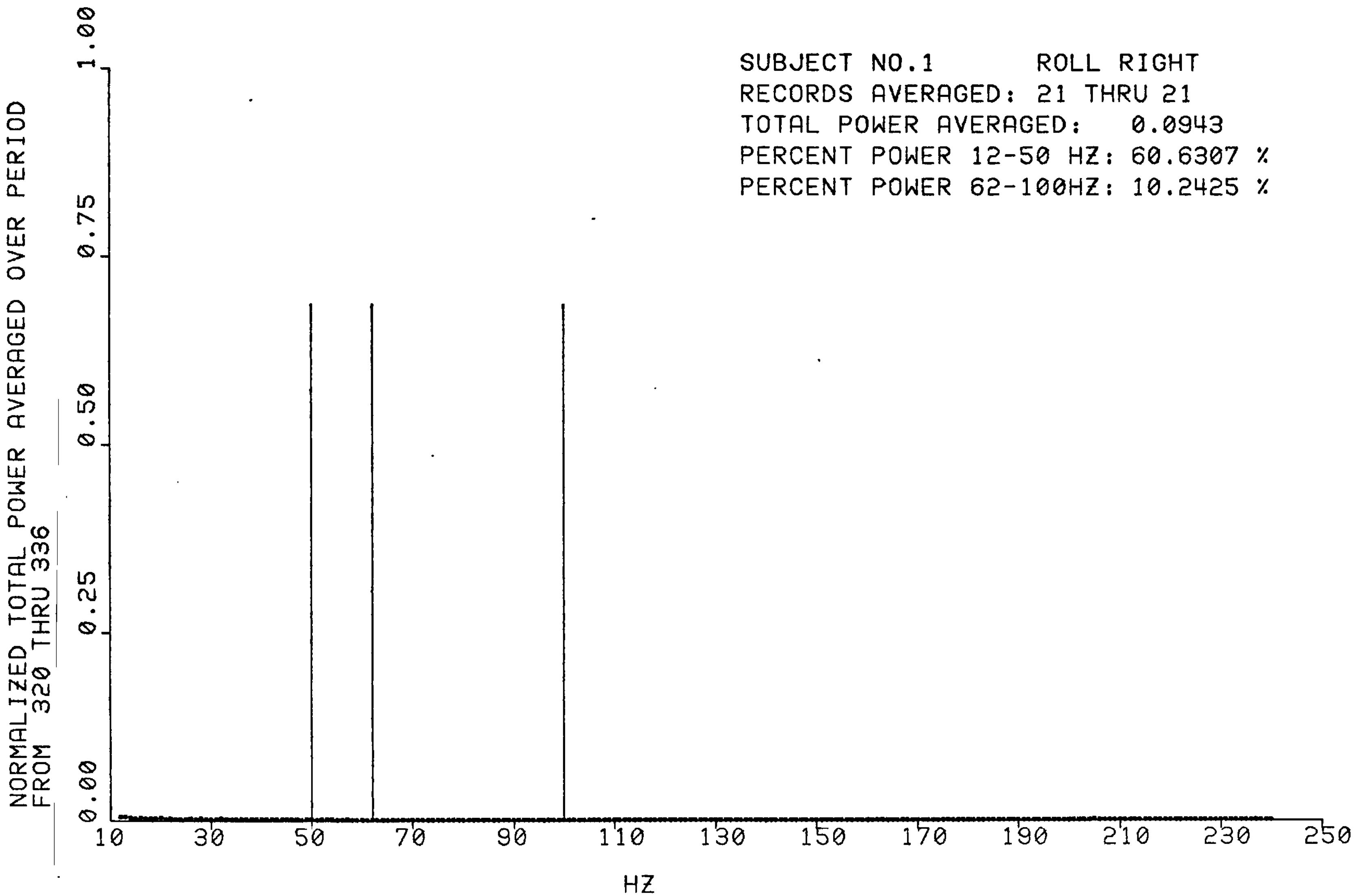

SUBJECT NO.1 ROLL RIGHT
RECORDS AVERAGED: 21 THRU 21
TOTAL POWER AVERAGED: 0.0943
PERCENT POWER 12-50 HZ: 60.6307 %
PERCENT POWER 62-100HZ: 10.2425 %
NORMALIZED TOTAL POWER AVERAGED OVER PERIOD
FROM 320 THRU 336
0.00
0.25
0.50
0.75
1.00
10
30
50
70
90
110
130
150
170
190
210
230
250
HZ

FIGURE A-15