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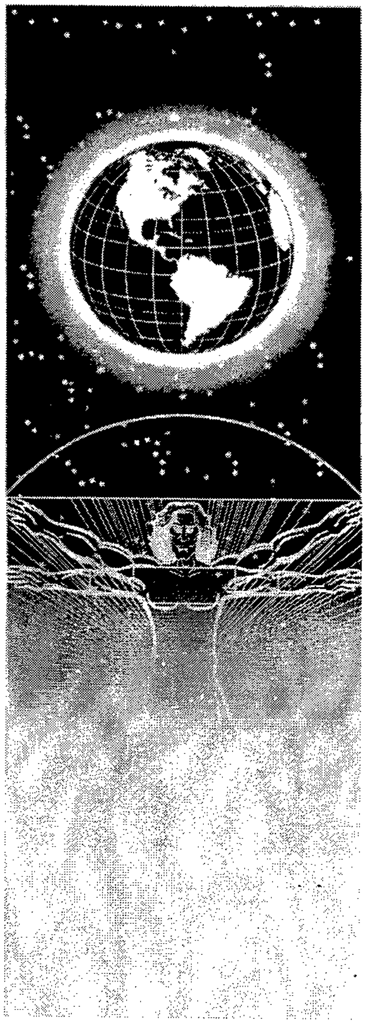

UNITED STATES AIR FORCE RESEARCH LABORATORY

Neck Muscle Response to Changes in Helmet Loading Under $+G_z$ Acceleration – Gender Differences

Edward S. Eveland

AIR FORCE RESEARCH LABORATORY

Chuck Goodyear

Statistical Consultant

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Human Effectiveness Directorate
Biosciences and Protection Division
Aircrew Protection Branch
2215 First Street
Wright-Patterson AFB OH 45433-7028

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14. ABSTRACT
Helmet-mounted devices (HMDs) can provide valuable assistance to the pilots of high performance aircraft. They allow operations to be accomplished in poor conditions and can improve effectiveness. However, each device changes the helmet configuration adding more weight and can change the center of gravity of the helmet. These factors are often implicated in reports of acute neck pain and cervical vertebral damage. Females, in particular, may be at risk for injury due to insufficient neck muscle strength to stabilize the neck under these loads. In this research, characteristics of the helmet configuration, along with +Gz acceleration, and task complexity are evaluated for their effect on human performance. Electromyography results showed that myoelectric activity of the neck muscles was most affected by the +Gz acceleration, whereas a helmet configuration effect was noticeable only for females. Other gender differences in myoelectric activity were also found. Females used as much as 80% of their maximal neck muscle capacity during performance of a task while males used 29% of their maximum. Approaching muscular limits might result in an inadequate muscular response to neck loading and the cervical area could be injured. In spite of this finding, tracking error of the females, although tending to be greater, was not significantly different from the error for males.

15. SUBJECT TERMS
Electromyography (EMG), myoelectric, simulated air combat maneuver, trapezius muscle, sternocleidomastoid muscle, magnetic resonance imaging (MRI), neck pain, neck muscle strength

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TECHNICAL REVIEW AND APPROVAL

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The voluntary informed consent of the subjects used in this research was obtained as required by Air Force Instruction 40-402.

This report has been reviewed by the Office of Public Affairs (PA) and is releasable to the National Technical Information Service (NTIS). At NTIS, it will be available to the general public, including foreign nations.

This technical report has been reviewed and is approved for publication.

FOR THE DIRECTOR

//Signed//

MARK M. HOFFMAN
Deputy Chief, Biosciences and Protection Division
Air Force Research Laboratory

PREFACE

This report was prepared in the Biodynamics and Acceleration Branch, Biodynamics and Protection Division, Human Effectiveness Directorate, of the Air Force Research Laboratory located at Wright-Patterson AFB, Ohio. The research was performed under work unit 71844501, Acceleration Performance in Advanced Operational Systems, with assistance from Veridian Engineering (contract # F41624-97-D-6004, Delivery Order # 022). The protocol was titled "Identification of Biodynamic Performance Parameters of Helmet Systems" (99-04, FW19990004H).

Thanks are due to the Dynamic Environment Simulator (DES) team for their professional approach to testing, leading to successful completion of the research effort. Technical assistance was provided by Per Bergman and Dr. Gianluca De Luca of Delsys Inc. and by Drs. Juhu Oksa and Olavi Hamalainen. Mr. Andy McKinley also deserves thanks for his assistance in the preparation of this report.

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INTRODUCTION

The potential for aircrew neck injuries is an Air Force concern that may become increasingly problematic as helmets are used as platforms for attachment of helmet-mounted devices (HMDs). These HMDs are valuable tools used in the performance of the mission as they can provide information that enables individuals to keep their attention outside the cockpit where potential threats can be more quickly ascertained. In addition, they allow crewmembers to be effective at night and in other adverse conditions. Still other HMDs provide sighting or triggering devices for new weapons. Unfortunately, these valuable tools come with a price. It becomes necessary for the neck to support the added weight of the extra equipment. The additional devices also tend to affect the weight distribution and alter the center of gravity (CG) of the helmet-HMD system. These changes in CG make it more difficult for the individual crewmember to support the head.

BACKGROUND

Many people have studied the effects of $+G_z$ acceleration on pilots. The high performance aircraft available today are capable of very high speed and maneuverability. The force experienced by pilots is dependent upon aircraft acceleration, direction, and inertial forces acting on the head and helmet. The G value for an acceleration is shown by G = applied acceleration/g where "g" is the gravitational constant with the value 9.81 m/s^2 (5). The z-axis (head to foot in the positive z direction) component of the G load (G_z) is of primary concern in this research due to its influence on the increased effective weight of an object under high G. For example, at 9 G, a four-pound helmet has an effective weight of 36 pounds. Relatively small cervical vertebrae and the neck muscles around them must support this increased load.

Modern aircraft maneuverability allows pilots to accelerate through turns with the feet and bottom of the aircraft along the outer radius of the curve. Inertial forces act in opposition to the acceleration and tend to exert a pull on the human body as illustrated in Figure 1. As aircraft maneuver and accelerate, both aircraft and pilot are affected by the acceleration illustrated by the bottom arrow. This acceleration moves the pilot up in space with the aircraft; however, there's an equal and opposite inertial force that tends to make any unattached part of the pilot feel like it drops. Blood tends to pool in the lower body.

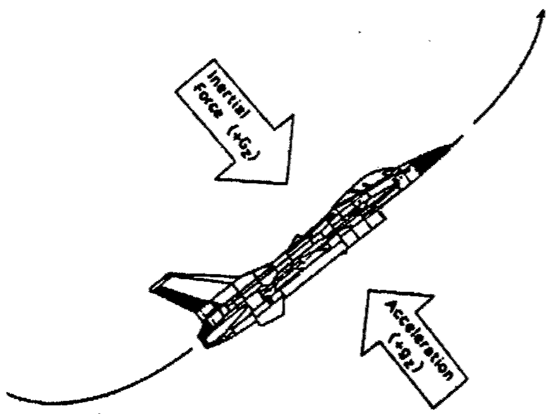


Figure 1. Maneuvering Acceleration and the Equal, but Opposite, Inertial Force that Aircrew are Exposed to During Flight (**3**)

Moments of inertia about the neck can also change, resulting in further stress as the CGs of the helmet systems change. The neck muscles must act to stabilize the neck. Unfortunately, the human element has not advanced as quickly as the aircraft. The cervical vertebrae in particular might be vulnerable if neck muscles are unable to properly support the loads. Vanderbeek (**12**) and others (**13**) suggest that the cervical vertebrae may be the next high-G "weak link" in the human system.

The research described in this report focuses on the forces acting upon the head and the support of the neck muscles forced to counter the inertial forces. If caught with their heads out of position at onset of acceleration, aircrew can literally have their heads pulled down as a result of the G load and be unable to lift them back to upright positions until the G is unloaded (**1**). Sometimes this can result in pain or injury to the neck.

Forces are directed through the vertebrae and compress the intervertebral disks (IVDs) evenly as long as the head remains directly over the spine and against the headrest. This allows the disks to act in the way they function best and to absorb the force. However, during high G_z exposure, the head can pitch forward, unable to resist inertial forces, especially if caught in an unfavorable position away from the normal "neutral" position. This forward movement of the head is usually observed when pilots move their head to look down at instruments or to look behind and "check six" for their adversary. Such actions make the crewmember more vulnerable to neck injury or pain due to the fact that the pilot must strain his/her neck extensor muscles to counteract the movement (7).

Surveys indicate that many high performance aircraft pilots experience in-flight neck pain. Some of those surveys show that greater than one-half of the respondents reported neck pain. The pain has generally been attributed to greater neck loading due to high-G maneuvers. It is interesting that most pilots who did not report experiences with neck pain took part in some type of exercise program. Hamalainen determined that "pilots are more susceptible to acute neck injury when protection from neck muscles is insufficient" (6). This recognizes the importance of neck muscles and the strength they provide.

Most injuries contributing to the neck pain pilots experience have been found in the cervical region of the spine. The cervical spine is most directly affected by weight of the head and helmet, which become heavier proportional to the G loads, and this part of the spine is most susceptible to injury (9). Common injuries include herniated or degenerated disks, compression fractures, cervical nerve damage, and ligament damage. Hamalainen studied Finnish fighter pilots using magnetic resonance imaging (MRI) to rate the severity of disk protrusion into the spinal canal. He found that the occurrence and the median degree of disk degeneration were greater among pilots than controls (8).

The greatest differences in degeneration were in the C3-C4 region. This finding differs from age-related cervical changes most commonly found in C5-C6. The conclusion based on that research was that frequent exposure to high $+G_z$ loads might cause premature cervical disk degeneration.

There are concerns that women now flying high performance aircraft for the Air Force or Navy may not have adequate neck strength to withstand high G encountered in the operational flight environment. Women generally do not have the muscularity of men, yet they must support the same helmets and HMDs. In research just being completed in the Air Force Research Laboratory (**2**), anthropometric measurements show larger neck circumferences for males than for females. The median values were 390 cm for the males but only 314 cm for the female participants. This may indicate less neck muscle availability. Cross-sectional diameter of muscle is important in strength. Smaller males may also be more susceptible to injury if they have weaker muscular support.

Neck injury leads to loss of flight time, which is undesirable for both the Air Force and the individual involved. While recovering from an injury, the Air Force is without a pilot, sometimes forcing others to do double duty and preventing the injured person from maintaining G tolerance. In order to retain the optimal capability to fly and pull high Gs, pilots must have regular training. Without that training, it is believed that tolerance to G decreases and pilots must spend additional time rebuilding it (**10**).

OBJECTIVES

In an effort to improve current conditions and eliminate some aircrew neck pain and injuries, a research plan was developed to evaluate neck muscles as they work to support the head and maintain it in a correct, safe position. Objectives of the research project, from the Air Force point of view, were to determine how performance changed as helmet system characteristics varied. An important part of the objective in this research was to better define the role neck muscles play in stabilizing the neck and head while supporting HMDs in an operational environment. That environment, in this case, is sustained acceleration. Characterization of neck muscle electrical activity and anything providing evidence that, when compared to males, females use a relatively larger proportion of their maximum muscular capability to stabilize their heads, resisting the strains attributed to high G acceleration should be valuable.

Without the neck strength of most men, women might find it necessary to strain more often or harder, causing early fatigue as they wear heavier helmet systems. Even during "routine" operations, smaller individuals have to withstand stress from helmets weighing essentially the same as helmets used by their larger male counterparts. In addition, this inferior helmet-bearing performance could result in inferior "helmet aiming/sighting" performance and compromise the mission. The purpose of this research was to find evidence allowing the rejection of the hypothesis and to determine if women might be at increased risk when the hypothesis is true. Information from this project may contribute to methods to reduce neck pain or injury and possibly identify potential problems with heavier helmet systems – especially during long duration missions in which muscle fatigue may be a factor.

METHODS

Ten qualified volunteer subjects from the sustained acceleration research panel participated in this research. The ten selected included five males and five females. All experimentation was conducted in the Dynamic Environment Simulator (DES) at Wright-Patterson AFB, OH. The DES (Figure 2) is a man-rated, three-axis centrifuge used to simulate the acceleration environment encountered by pilots.

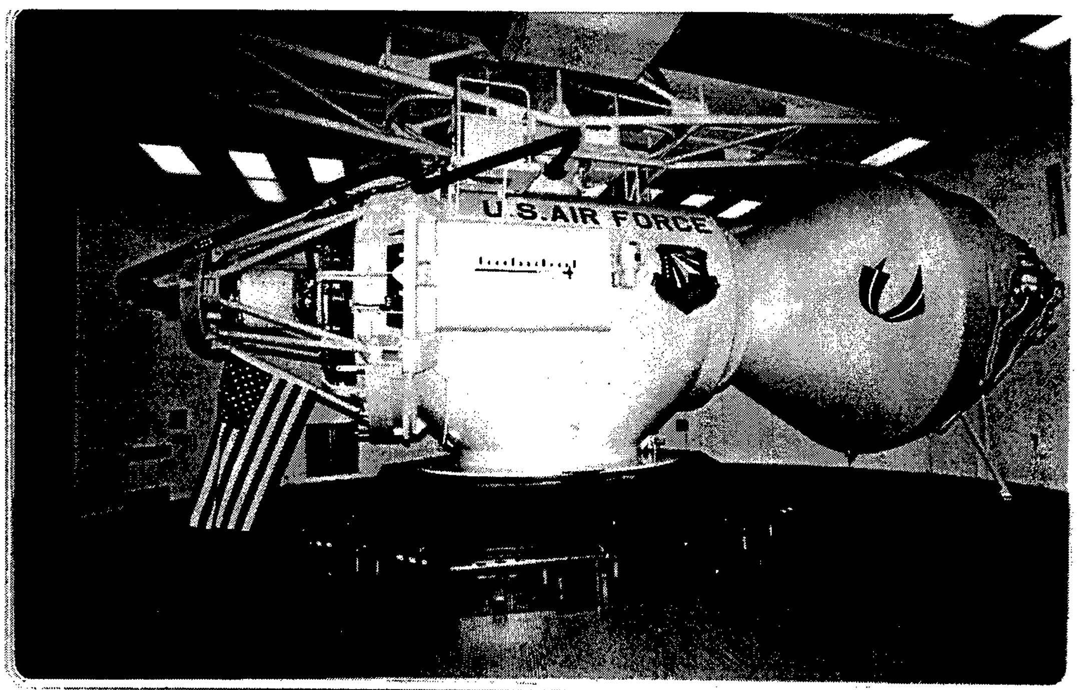


Figure 2. The Dynamic Environment Simulator (DES)

Standard Air Force HGU-55/P helmets were custom-modified for this experimentation (Figure 3). The modifications allowed weights to be attached and secured along a wire rail system. The weights were positioned anywhere from over the ears to the front brow area of the helmet. This method permitted adjustments to the center of gravity depending on positioning and amount of attached weights. Weights placed in the most forward position could be expected to result in a forward movement in the CG. A laser pointer, for tracking, was mounted on the top of the helmet and extended down so that the laser was located in a position approximately between the eyes of the wearer. The position was adjustable.

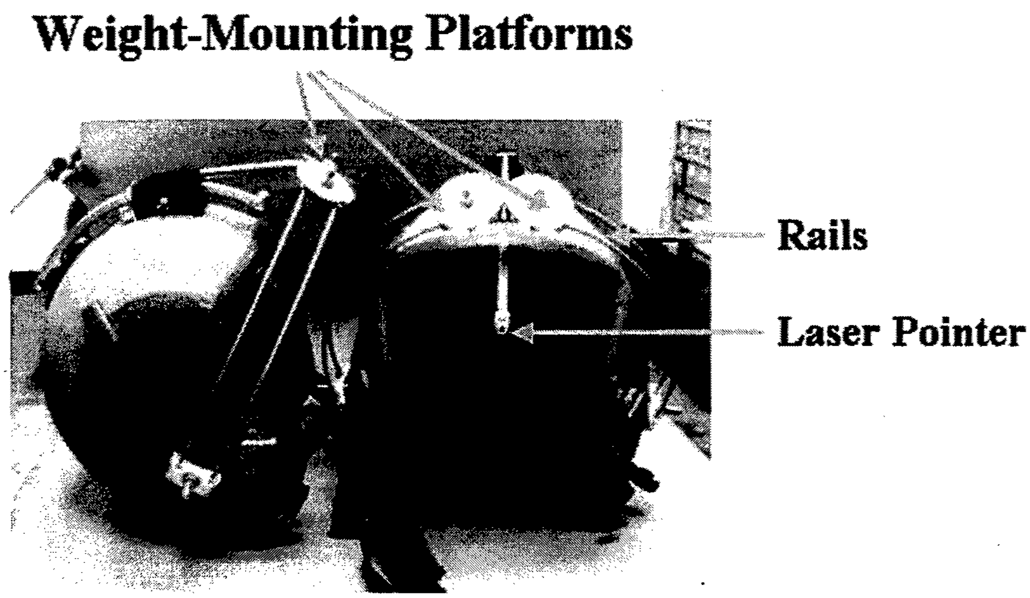


Figure 3. Custom-Modified HGU-55/P Helmet with Laser Pointer

All subjects completed testing on six separate visits or experimental days at the DES facility. There were two runs per day with a 10-minute break between. On any given day, an individual wore a helmet with or without added weight. The same helmet weight was used for both runs that day. CG was adjusted as weight was repositioned during the break when the heavier helmet was used. During the runs, the subjects were required to perform a tracking task. They were asked to track a moving target by moving the helmet-mounted laser pointer during task periods. The goal was to keep the red dot from the laser pointer on the moving target. The target was set to move at different speeds depending on which task was running. The simple task was slower than the complex task.

The lightest helmet was 2.5 pounds and designated 2.5-0. It was made heavier on four of the six days by adding one-pound of weight. In this notation, used to describe helmet configuration, a "0" after helmet weight designates the neutral position (2.5-0 or 3.5-0) and a "1" after the weight designates the configuration with half of the one pound total positioned forward on the helmet for an extended CG (3.5-1). The CG of the helmet would be moved forward with the weight in this second position. This one-pound weight was added either all near the ears for a more neutral center of gravity (3.5-0) or split so one-half pound was placed at the ear location and half placed forward, toward the front of the helmet (3.5-1). Presentation of task and helmet configuration was counterbalanced within each subject to minimize order effects.

In the first configuration, with a half pound forward, the center of gravity along the Z-axis for the large helmet was at 1.01 inches, and along the X-axis, it was at 0.11 inch. For the second configuration, all the weight was placed in the ear position. The center of gravity along the Z-axis (CGZ) was at 0.9 inch, and along the X-axis (CGX), it was located at –0.17 inch.

Electrical Recording (Electromyography)

Two muscle pairs were instrumented to collect electrical signals indicative of muscular activity. The sternocleidomastoid (SCM) and trapezius (T_z) muscles were set up for measurement of myoelectric activity. DelSys DE-2.1 single differential surface electrodes (4) were attached with the casing parallel to expected muscle fiber orientation using electrode interfaces provided by the manufacturer. These interfaces are double-sided sticky materials using medical grade adhesives designed to stick tenaciously even in environments of high humidity during vigorous activity. They are also said to maintain contact stability and reduce motion artifacts. The electrodes are the standard electrodes supplied with the Bagnoli-4 electromyography (EMG) system (4). The electrode housing contained both metal contacts so the distance between electrode pairs was controlled and consistent for each test.

Since skin resistance is very important in allowing the miniature currents from muscle to reach the electrodes, the skin at the contact points was prepared prior to application by scrubbing with isopropyl alcohol and drying with gauze pads. Very small amounts of conducting gel were applied to the electrode contact points (bars), using care to avoid excess gel that would allow a conductance path between contacts. The reference electrode was a standard EKG electrode applied to the skin over the 7th cervical vertebra. Real-time data collection used in this research allowed the investigators to see graphic representations of the muscle activity as it occurred. Any problems could be corrected before the trials began. Four channels of muscle activity data were displayed with one additional channel indicating the level of G exposure (Figure 4). This allowed researchers to watch and later identify muscular response to the added challenge of high G acceleration. It was possible to see real-time power spectrum and median frequency displays during the test runs.

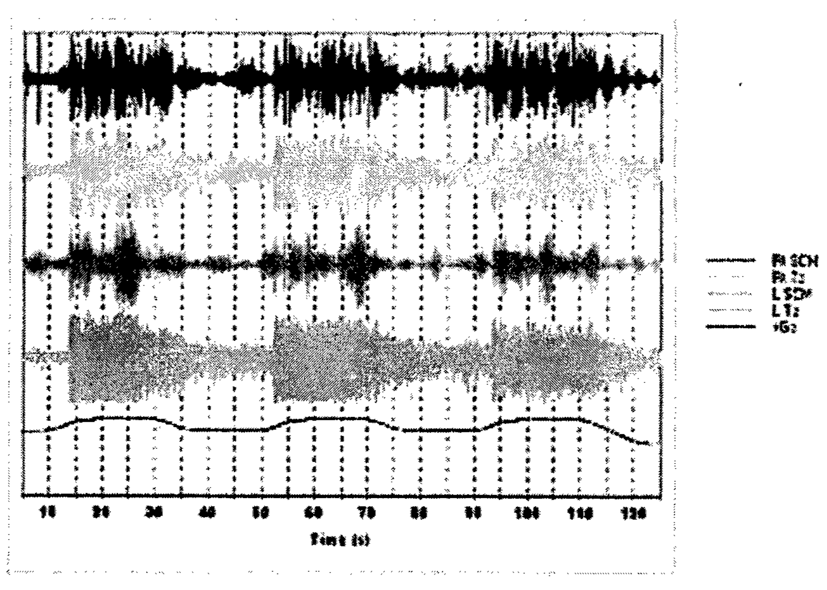


Figure 4. Electromyography Data Collection Display (4)

Immediately prior to acceleration testing on each experimental day, subjects were asked to provide measures of the maximum capability of their neck muscles in extension. They were instructed to perform three consecutive maximum voluntary contractions

(MVCs). While seated, a head harness used in neck muscle strength training was attached to the head (Figure 5) and the fit adjusted with hook-and-loop tabs. A chain ran from the harness to a mounting bracket near the feet. The attachment point of the bracket was adjusted so the subject could pull in an upright position with a head angle similar to the head angle the individual assumed during tracking.

Figure 5. Testing for Maximal Voluntary Contractions (MVCs)

With electromyography data collection software running, subjects pulled with maximal effort for three attempts lasting 3-5 seconds each. EMG tracings were displayed on the screen where subjects could see them for motivation. They served to provide positive feedback. Each attempt was separated by a short rest period. Upon completion of MVC testing, the harness was removed and the subjects donned a helmet for the next phase. A second set of MVC attempts was done following completion of all the experimentation for the day. Making the second measurement provided an opportunity to compare pre- and post-test values to see if subjects' maximal capability dropped after a day of strenuous exertion.

In anticipation of needing force data, some of the final MVC sessions were conducted with instrumentation connected in series within the chain line-of-pull. All procedures were the same in these MVC tests except for these measurements.

Experimental testing was based on the concept diagram in Figure 6. All testing for any one day followed the pattern, from blocks 1-4, in that illustration. After MVC testing (Block 1), subjects remained in the cab of the DES centrifuge and prepared for exposure to experimental variables. They encountered two types or levels of sustained acceleration: (1) a 1.4 G_z baseline condition, and (2) a 6.5 G_z peak SACM (Simulated Air Combat Maneuver) for approximately 120 seconds each. A SACM simulates an aerial dogfight in combat with two adversaries approaching, passing, and then turning to re-

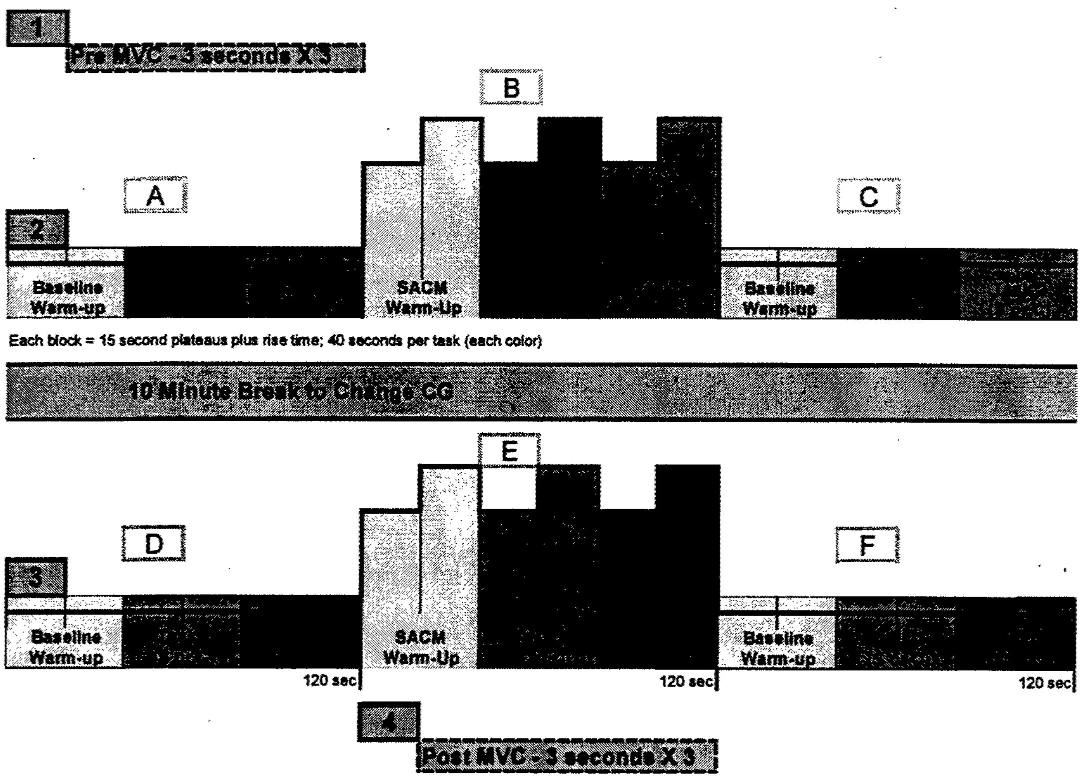


Figure 6. Experimental Concept Diagram

engage the enemy. In this case, the SACM included a G profile including both 4 and 6.5 G plateaus (Figure 7). A computer was programmed to provide the subject with a slow onset exposure to 4 G, where they remained for 15 seconds, and then to immediately increase the speed of the DES to provide 6.5 G_z for another 15-second interval. Again, upon completion, another set of 4 to 6.5 G plateaus would begin. Three sets of 4 to 6.5 G_z plateaus were included in a single SACM.

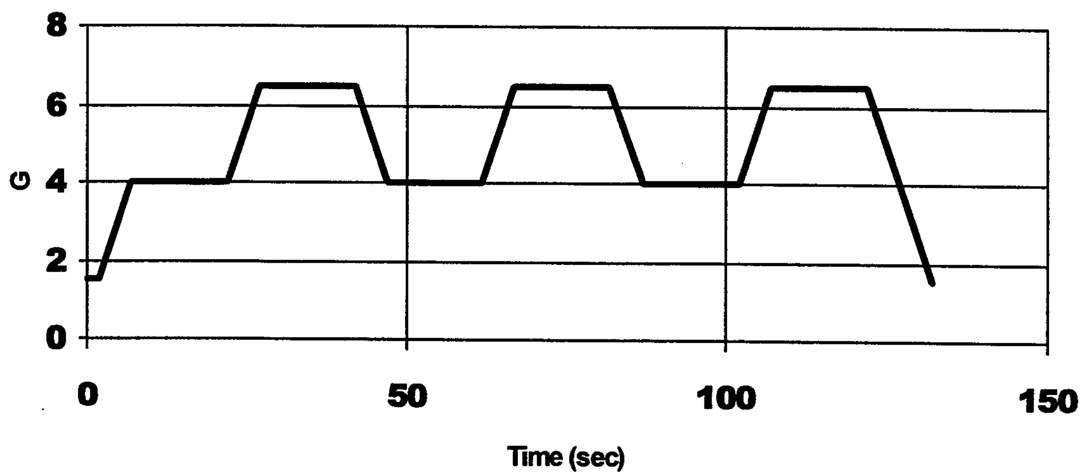


Figure 7. Acceleration Profile

The subject performed one set of three runs with the helmet configured for one CG position. The set of runs included the baseline G_z condition first. This is illustrated by "A" in Figure 6. A high-G challenge using the SACM was presented next in "B" and followed by a post-challenge baseline "C". In each of these segments on the chart or actual runs of the centrifuge, there was an initial warm-up period and two task periods. The task order changed so that the subject did not always experience either the Simple (S) or the Complex (C) task first. During a ten-minute break after the first set of runs (A, B, C), technicians re-entered the centrifuge cab and redistributed weights. The subject then performed three more runs (D, E, F) using the new helmet CG position as shown in Table 1. The order was counterbalanced. Performance on the target tracking tasks was evaluated throughout all parts of the G exposures.

Table 1. Experimental Design. The row entry under each subject column is the experimental day. The run code identifies the helmet weight, CG, and task order used on that run. S = Simple and C = Complex tracking tasks.

Run Code		Run	Day Order for each Subject									
			Female Subject #					Male Subject #				
			3	5	6	15	25	7	18	19	20	22
2.5	0 SC 0 CS	1 2	5	1	6	3	2	4	6	1	5	3
2.5	0 CS 0 SC	1 2	2	5	3	6	4	1	3	4	2	6
3.5	0 SC 1 CS	1 2	1	6	2	4	3	2	4	5	3	1
3.5	1 SC 0 CS	1 2	4	3	5	1	6	5	1	2	6	4
3.5	1 CS 0 SC	1 2	3	4	1	2	5	3	2	6	1	5
3.5	0 CS 1 SC	1 2	6	2	4	5	1	6	5	3	4	2

Numbers at the left in the table denote the helmet weight. The weight was either 2.5 or 3.5 pounds. SC or CS indicates the order of tracking tasks, SC = Simple/Complex and CS = Complex/Simple, over trials 1 and 2 which follow. The next column, zero or one, indicates weight position. A "0" denotes the neutral position near the ears and a "1" represents the configuration with half the weight placed forward. The order in which any individual subject encountered a particular combination of helmet weight and task order is shown under "Run Code" for any particular visit listed under "Day Order for each Subject". For example, female subject number 15, on day 4 (beneath subject #), was scheduled to wear the 3.5-lb. helmet. She had run #1 with the extra weight mounted in a neutral position and a SC task order, so she experienced the Simple task first. This helmet configuration is referred to in this research as 3.5-0. In run #2, she had half the weight moved forward, changing the CG (3.5-1). Task order was reversed to CS. Since there is no weight to move on the lightest helmet, there is no CG change and therefore fewer runs with the lighter helmet weight.

RESULTS

Repeated measures analysis of variance results are shown in shown in Tables 2-5.

Factors were gender (male, female), helmet weight – CG (2.5-0, 3.5-0, 3.5-1), task (Simple, Complex), and Gz (1.4, 4.0, 6.5). Post hoc paired comparisons used 2-tailed t-tests with pooled error. The effect of helmet weight was determined by comparing helmets 2.5-0 and 3.5-0, while the effect of center of gravity was determined by comparing helmets 3.5-0 and 3.5-1. All tests using Mean Absolute Value (MAV) were performed on log-transformed percents; however, all tables and figures express means in percents. Significant results are shaded.

Table 2. Analysis of Variance Results. The dependent variable was the log of percent of pre-MVC. Variable = MAV, Muscle = SCM.

Source	DF	SS	F	p	G-G p	G-G Ep
Gender (G)	1,8	8.78E+00	8.74	0.0183		
Helmet (H)	2,16	1.26E-01	4.08	0.0369	0.0700	0.5726
Task (T)	1,8	2.66E-02	4.57	0.0649		
Gz (Gz)	2,16	1.54E+01	198.50	0.0001	0.0001	0.7268
G*H	2,16	1.68E-01	5.43	0.0159	0.0409	0.5726
G*T	1,8	3.75E-03	0.65	0.4450		
G*Gz	2,16	6.05E-02	0.78	0.4753	0.4423	0.7268
H*T	2,16	3.22E-03	1.19	0.3307	0.3264	0.8345
H*Gz	4,32	5.28E-03	0.59	0.6726	0.6141	0.6860
T*Gz	2,16	2.97E-02	7.85	0.0042	0.0047	0.9668
G*H*T	2,16	5.20E-03	1.92	0.1794	0.1882	0.8345
G*H*Gz	4,32	8.82E-03	0.99	0.4295	0.4119	0.6860
G*T*Gz	2,16	9.74E-04	0.26	0.7764	0.7693	0.9668
H*T*Gz	4,32	1.51E-03	0.54	0.7092	0.6025	0.5255
G*H*T*Gz	4,32	3.58E-03	1.27	0.3021	0.3079	0.5255

DF = degrees of freedom for source, error term
SS = source sum of squares
F = F-value
p = p-value
G-G p = p-value using the Greenhouse-Geisser correction
G-G Ep = epsilon value from Greenhouse-Geisser correction

Table 3. Analysis of Variance Results. The dependent variable was the log of percent of pre-MVC. Variable = MAV, Muscle = T_Z.

Source	DF	SS	F	p	G-G p	G-G Ep
Gender (G)	1,8	8.71E+00	16.03	0.0039		
Helmet (H)	2,16	8.16E-04	0.02	0.9838	0.9210	0.5592
Task (T)	1,8	4.04E-03	1.53	0.2505		
Gz (Gz)	2,16	1.06E+01	37.22	0.0001	0.0002	0.5182
G*H	2,16	2.78E-02	0.56	0.5830	0.4932	0.5592
G*T	1,8	4.62E-06	0.00	0.9676		
G*Gz	2,16	2.58E-02	0.09	0.9142	0.7801	0.5182
H*T	2,16	1.05E-03	0.90	0.4247	0.4095	0.8185
H*Gz	4,32	4.88E-03	0.55	0.6974	0.5843	0.4972
T*Gz	2,16	1.49E-02	1.22	0.3215	0.3133	0.6941
G*H*T	2,16	1.39E-03	1.19	0.3293	0.3246	0.8185
G*H*Gz	4,32	8.91E-03	1.01	0.4163	0.3856	0.4972
G*T*Gz	2,16	4.29E-03	0.35	0.7089	0.6352	0.6941
H*T*Gz	4,32	5.50E-03	1.58	0.2043	0.2279	0.6454
G*H*T*Gz	4,32	6.82E-03	1.96	0.1253	0.1582	0.6454

Table 4. Analysis of Variance Results. The dependent variable was the percent of pre-MVC. Variable = MDF (Median Frequency), Muscle = SCM.

Source	DF	SS	F	p	G-G p	G-G Ep
Gender (G)	1,8	6.43E+03	1.91	0.2039		
Helmet (H)	2,16	2.93E+02	0.37	0.6994	0.5717	0.5251
Task (T)	1,8	2.02E+01	0.49	0.5038		
Gz (Gz)	2,16	8.47E+02	1.20	0.3273	0.3163	0.6597
G*H	2,16	1.56E+02	0.19	0.8250	0.6822	0.5251
G*T	1,8	3.31E+01	0.80	0.3963		
G*Gz	2,16	6.95E+02	0.98	0.3955	0.3697	0.6597
H*T	2,16	1.67E+01	0.16	0.8517	0.7489	0.6209
H*Gz	4,32	4.23E+02	2.13	0.1002	0.1371	0.6207
T*Gz	2,16	2.49E+02	7.11	0.0062	0.0174	0.6601
G*H*T	2,16	1.47E+01	0.14	0.8680	0.7667	0.6209
G*H*Gz	4,32	3.39E+02	1.71	0.1729	0.2030	0.6207
G*T*Gz	2,16	4.76E+01	1.36	0.2854	0.2833	0.6601
H*T*Gz	4,32	7.42E+01	1.02	0.4143	0.3787	0.4473
G*H*T*Gz	4,32	1.92E+02	2.63	0.0526	0.1107	0.4473

Table 5. Analysis of Variance Results. The dependent variable was the percent of pre-MVC. Variable = MDF, Muscle = T_Z.

Source	DF	SS	F	p	G-G p	G-G Ep
Gender (G)	1,8	3.00E+03	0.39	0.5480		
Helmet (H)	2,16	4.00E+02	0.68	0.5220	0.4374	0.5121
Task (T)	1,8	1.90E+02	12.60	0.0075		
Gz (Gz)	2,16	1.15E+03	1.04	0.3755	0.3525	0.6425
G*H	2,16	2.01E+03	3.41	0.0585	0.1007	0.5121
G*T	1,8	1.37E+02	9.12	0.0166		
G*Gz	2,16	3.69E+02	0.34	0.7201	0.6296	0.6425
H*T	2,16	2.16E+01	0.33	0.7233	0.6371	0.6572
H*Gz	4,32	1.47E+02	0.93	0.4578	0.4142	0.5003
T*Gz	2,16	2.40E+01	0.62	0.5529	0.5139	0.7605
G*H*T	2,16	9.04E+01	1.38	0.2789	0.2781	0.6572
G*H*Gz	4,32	7.59E+01	0.48	0.7498	0.6272	0.5003
G*T*Gz	2,16	1.21E+01	0.31	0.7385	0.6815	0.7605
H*T*Gz	4,32	8.19E+01	0.92	0.4644	0.4305	0.5930
G*H*T*Gz	4,32	1.02E+02	1.14	0.3543	0.3479	0.5930

Figures 8-11 follow the analysis of variance tables and illustrate the results of those analyses. The figures contain main effect means and simple main effect means from significant interactions. Since percents for MAV were logged for analysis, the Y-axis on the figures is on a log scale. Means on top of the bar graphs are geometric means. Post hoc test results are provided when relevant.

Figure 8 depicts the significant gender differences from Table 2. Females are shown to use 80 percent of their pre-MVC amplitude while males used only 29 percent of their pre-MVC values. Acceleration effect (G_z) was also shown to be significantly different. The plot on the right end of the figure shows 19 percent used at baseline, then an increase to 60 percent at 4.0 G_z and still higher values (97 percent) at +6.5 G_z.

Significant interactions are shown in the final two plots of Figure 8. No significant differences were found for the Gender*Helmet interaction for either female (p = 0.0904) or male (p = 0.1030) subjects. There were, however, significant differences between females and males for helmet 2.5-0 (p = 0.0294), 3.5-0 (0.0153), and 3.5-1 (p = 0.0172). All means from the females were much larger percentages of pre-MVC than the means from the males.

Simple main effect F-tests showed significant differences for the interaction of task and G_z ($Task*G_z$) in both the Simple ($p = 0.0001$) and Complex (0.0001) tasks. T-tests showed each level of G_z significantly different from the others ($p \leq 0.0001$) for both tasks. Simple main effect F-tests showed a significant difference between the tasks for G_z=1.4 ($p = 0.0033$), but not for G_z=4.0 ($p = 0.1346$) or for G_z=6.5 ($p = 0.4325$).

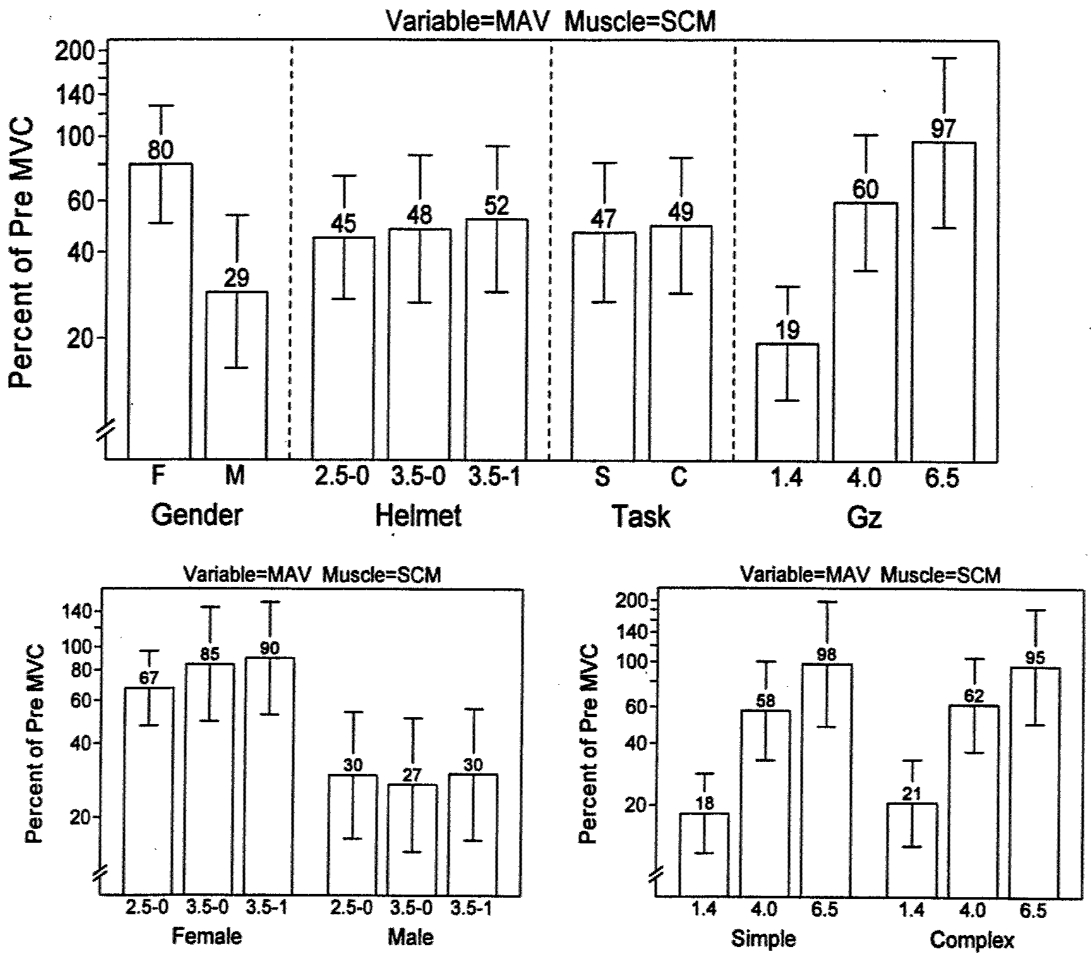


Figure 8. Main Effect Means and Simple Main Effect Means from Significant Interactions ± Standard Deviation of Subjects

Amplitude of the trapezius muscle EMG again showed a large gender difference (Table 3, Figure 9). In this case, the ratio was 55 percent for the females in our research sample and 20 percent for the males. The second significant difference was related to acceleration and increased in a stepwise manner from baseline to 4.0 G and then 6.5 G.

T-tests for the G_z main effect means showed each level of G_z significantly different from the others ($p \leq 0.0067$).

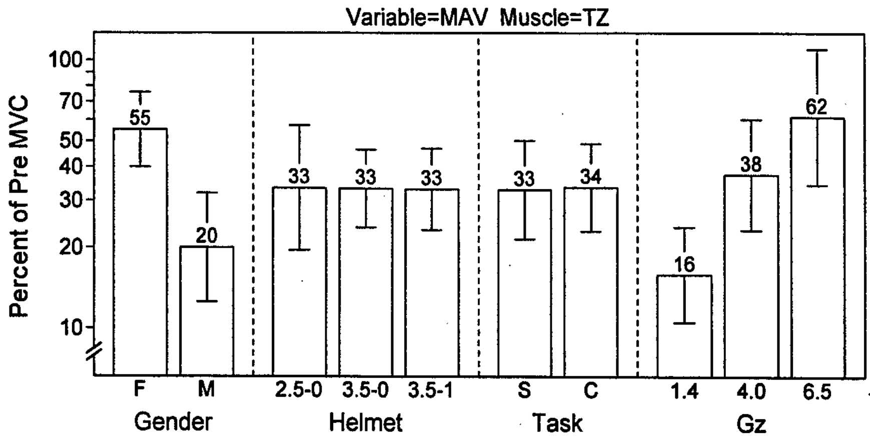


Figure 9. Main Effect Means ± Standard Deviation of Subjects

There was only one significant difference shown in analysis of the MDF data from the SCM (Table 4). It was an interaction between task and G level (T^*G_z). For this Task*G_z interaction (Figure 10), simple main effect F-tests did not show a significant difference among the G_z levels for either the Simple ($p = 0.7188$) or Complex ($p = 0.1015$) task. Simple main effect F-tests showed a significant difference between the tasks for G_z=6.5 ($p = 0.0254$) but not for G_z=1.4 ($p = 0.0984$) or G_z=4.0 ($p = 0.3629$).

The final table (Table 5) showed the significant differences from analyses of frequency data collected from the T_z. The percent of pre-MVC for MDF was higher for Simple tasks than for Complex tasks (Figure 11). Muscle activity during Simple tasks fired at a rate equal to 106 percent of the rate measured in MVC testing prior to testing. The rate associated with the Complex task was lower, at 104 percent of Pre-MVC.

For the Gender*Task interaction, simple main effect F-tests did not show a significant difference between the tasks for the female subjects ($p = 0.6259$), but did show a significant difference between the tasks for the male subjects ($p = 0.0191$). Simple main effect F-tests did not show a significant difference between females and males for the Simple ($p = 0.4818$) or Complex ($p = 0.6245$) task.

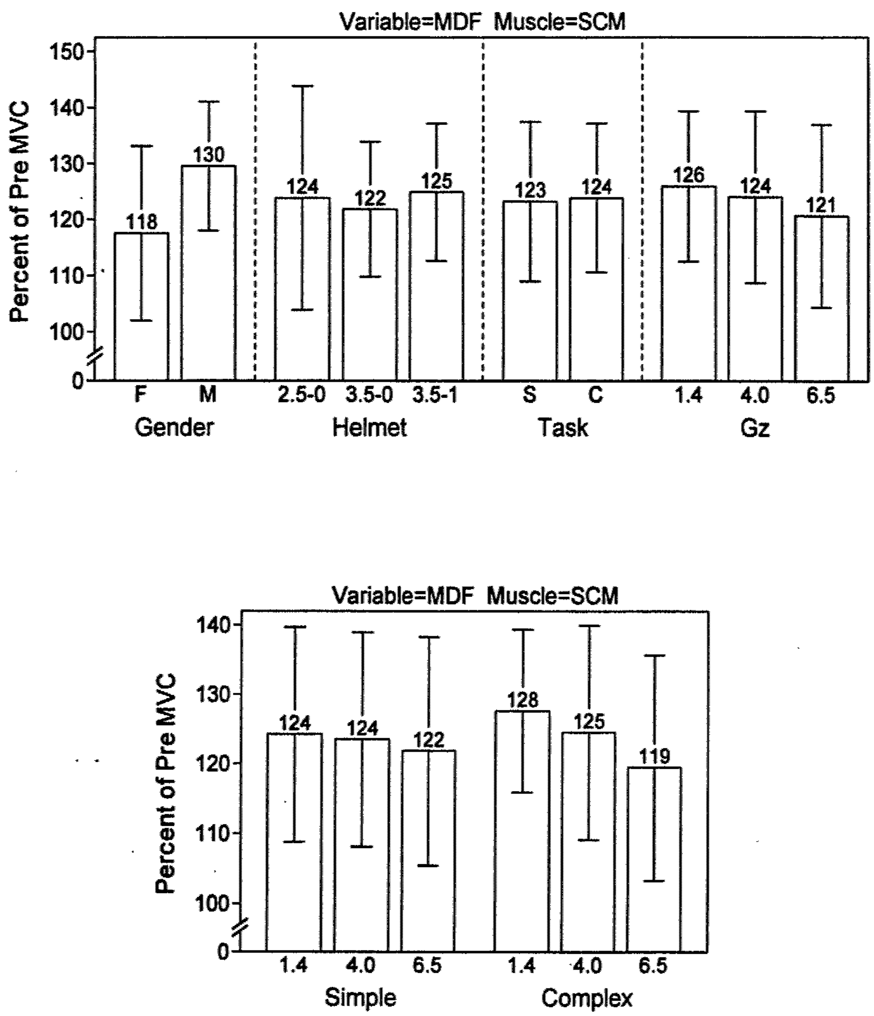
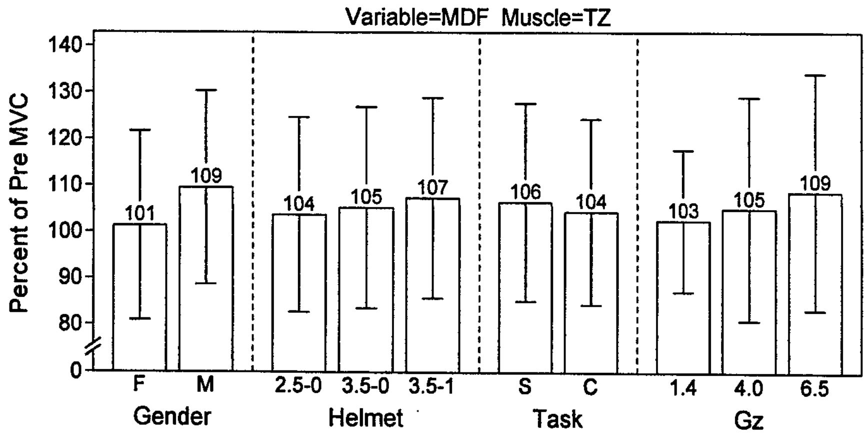


Figure 10. Main Effect Means and Simple Main Effect Means from the Significant Task*G_z Interaction ± Standard Deviation of Subjects



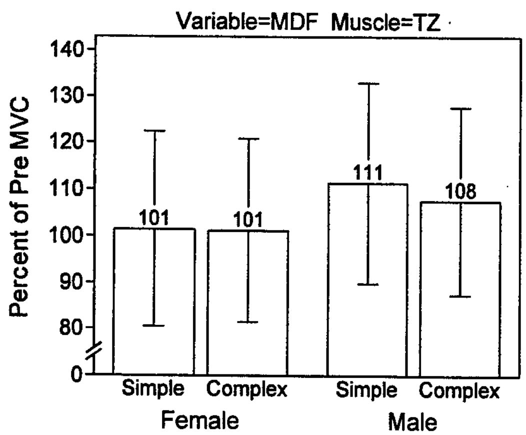


Figure 11. Main Effect Means and Simple Main Effect Means from the Significant Gender*Task Interaction ± Standard Deviation of Subjects

Tracking Performance

The root mean square (RMS) of tracking error was determined for each subject (1-10), day (1-6), segment (A-F), and task (Simple and Complex). The purpose of this analysis was to determine if gender, helmet weight-CG, task, or G level influenced performance as shown by RMS error.

RMS error values were positively skewed. To help normalize these RMS errors, a log transformation was used. For each combination of subject, helmet, and task, there were 16 replications for 1.4 G_z (pre and post together) and four replications for 4.0 G_z and 6.5 G_z. The mean of these replications was used as the dependent variable in a mixed design analysis of variance with gender (female and male) a between-factor and including within-factors helmet weight-CG (2.5-0, 3.5-0, and 3.5-1), task (Simple and Complex), and G_z (1.4, 4.0, and 6.5). The helmet weight - CG factor is referred to simply as helmet. Results are shown in Table 6. The Greenhouse-Geisser adjustment was used for all F-tests, including simple F-tests performed within significant interactions.

No significant gender differences were found in this analysis. Helmet, task, and G_z, however, were all found to be different. Post hoc paired comparisons used 2-tailed t-tests with pooled error. The effect of helmet weight was determined by comparing helmets 2.5-0 and 3.5-0, while the effect of center of gravity was determined by comparing helmets 3.5-0 and 3.5-1. All tests were performed on the log transformed RMS errors. Figure 12 uses a log scale on the Y-axis, while the means at each tick mark and on top of the bars are geometric means.

Table 6. Analysis of Variance Results. The dependent variable was the log of RMS error.

Source	DF	SS	DFE	SSE	F-value	P-value	G-G P-value	G-G Epsilon
Gender (G)	1	2.39E-01	8	1.30E+00	1.47	0.2599		
Helmet (H)	2	1.23E-02	16	2.38E-02	4.15	0.0353	0.0403	0.9131
Task (T)	1	1.36E+00	8	1.61E-02	678.75	0.0001		
Gz (Gz)	2	2.18E-01	16	1.79E-01	9.75	0.0017	0.0036	0.8222
G*H	2	5.75E-03	16	2.38E-02	1.94	0.1766	0.1814	0.9131
G*T	1	1.98E-04	8	1.61E-02	0.10	0.7614		
G*Gz	2	4.10E-02	16	1.79E-01	1.84	0.1915	0.2001	0.8222
H*T	2	2.87E-03	16	1.72E-02	1.33	0.2919	0.2908	0.7842
H*Gz	4	7.23E-03	32	2.40E-02	2.41	0.0696	0.0941	0.7275
T*Gz	2	1.75E-02	16	1.36E-02	10.28	0.0013	0.0034	0.7901
G*H*T	2	4.29E-03	16	1.72E-02	1.99	0.1688	0.1811	0.7842
G*H*Gz	4	5.99E-03	32	2.40E-02	2.00	0.1184	0.1433	0.7275
G*T*Gz	2	2.74E-03	16	1.36E-02	1.61	0.2312	0.2372	0.7901
H*T*Gz	4	1.58E-03	32	2.09E-02	0.60	0.6630	0.6041	0.6785
G*H*T*Gz	4	1.86E-03	32	2.09E-02	0.71	0.5892	0.5417	0.6785

T-tests for the helmet main effect means showed 2.5-0 significantly different from 3.5-1 ($p = 0.0138$), but showed no significant difference between 2.5-0 and 3.5-0 ($p = 0.5027$) or between 3.5-0 and 3.5-1 ($p = 0.0539$).

For the Task*G_z interaction (Figure 12), simple main effect F-tests showed significant differences among the G_z levels for both the Simple ($p = 0.0006$) and Complex ($p = 0.0360$) tasks. For the Simple task, t-tests showed G_z=1.4 significantly different from G_z=4.0 ($p = 0.0080$), but showed no significant difference between G_z=1.4 and G_z=6.5 ($p = 0.0621$) or between G_z=4.0 and G_z=6.5 ($p = 0.3223$). For the Complex task, t-tests showed G_z=1.4 significantly different from G_z=4.0 ($p = 0.0001$) and G_z=6.5 ($p = 0.0018$), but showed no significant difference between G_z=4.0 and G_z=6.5 ($p = 0.1444$). Simple main effect F-tests showed a significant difference between the tasks at each G_z level ($p \leq 0.0001$).

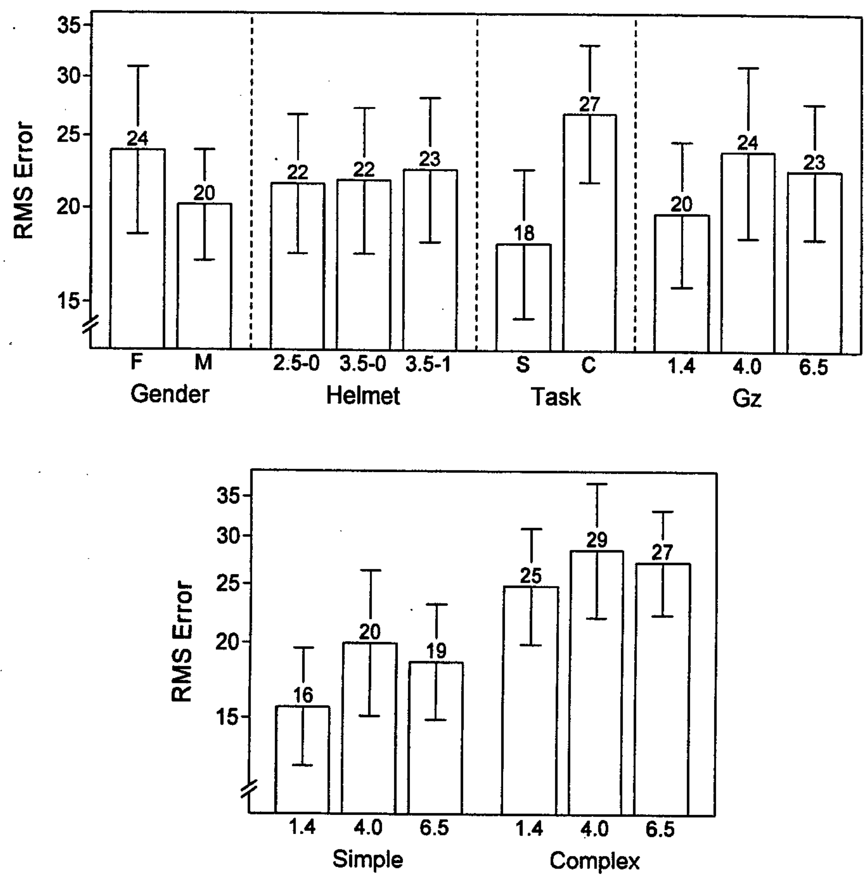


Figure 12. Main Effect Means and Simple Main Effect Means from the Significant Task*G_z Interaction ± Standard Deviation of Subjects

DISCUSSION

The net gain expected from this research was information describing how factors related to helmet system characteristics affect neck muscular activity and human performance in a high-G environment. Myoelectric activity did change in response to the challenges presented in this report. The research objective was to better define aircrew neck muscle activity while supporting helmet-mounted systems in an operational environment and to gather information that may contribute to design criteria for HMDs. Two of the helmet factors typically expected to be most important are weight and CG. Designers of HMDs could benefit from new information detailing how these factors affect pilots as they develop new systems and modify existing ones. Eventually, alleviation of neck pain and injury for the aircrew of the future might be possible.

Once data collection began, more specific information on gender was found. Increases in EMG signal amplitude and shifts in the power spectrum are indicators that could provide evidence of fatigue after exposure to test conditions. Changes in these EMG features are known to be common in fatigue situations (**11**). It now appears that females, in this research sample, are affected differently or at least to a greater degree than their male counterparts are affected.

The first evidence comes from an analysis of variance on amplitude data (MAV) from SCM muscles. This muscle group showed gender differences in the myoelectric response (Table 2). The main effect analyzed by gender was found to be significantly different as was gender interaction with helmet (G*H). Signal amplitude, usually indicative of motor unit recruitment, was a much larger percentage of pre-test MVC amplitude for females than for males after exposure to the challenge presented in this research. The ratio was an unbalanced 80 percent for females to 29 percent for the males (Figure 8).

Gender also played a role through its interaction with helmet configuration. The normalized MAV of amplitude data from the SCM for female subjects was a very high percentage of pre-MVC with all helmet configurations. Males showed no significant differences between any of the helmet configurations. Their myoelectric activity was much lower with all configurations. Females conversely used most of their neck muscular capability regardless of the HMD, increasing from the lightest helmet to the 3.5-0 helmet, and then the 3.5-1 version.

Thus, examination of the G*H interaction showed that, while males showed no differences in effect based on helmet configuration, females were greatly affected. When averaged over all G levels, the percent MVC for females, compared to males, was much greater for all helmets - the lightest 2.5 lb. helmet and for both of the heavier helmet configurations (3.5-0 and 3.5-1). The differences for the lighter helmet were somewhat smaller but were still double those values found for male subjects. These are key findings as helmet weight and CG location were expected to be critical to design criteria. Indications from this research are that high G acceleration is such a very large stress and influence on neck muscle activity of the females that it requires nearly maximal effort with any helmet. It remains to be seen if males might be affected similarly by helmet configuration when increased levels of muscular activity are required.

Loads due to G affected myoelectric signal amplitude, as one might expect. Subjects recruited more muscle against the stronger inertial forces in a stepwise manner with accelerations from baseline (1.4 G) to 4.0 and 6.5 G. Increased muscular support was needed to counter the greater forces occurring at higher G levels. Normalized muscle activity shown reached nearly maximum at 6.5 G_Z. The mean value was 97 percent. Each level of G (baseline, 4.0, and 6.5) was significantly different from the others.

G_Z, too, was found to have a significant interaction with another factor - this time with task complexity (T). When task was considered, there were differences at baseline. More activity, shown by a higher percent of pre-MVC, was found to occur in performance of the Complex tasks at baseline. Although significant differences were not found at higher G, Complex tasks at baseline resulted in significantly greater amplitudes (MAV values) as normalized percent of pre-MVC.

All these findings appear to support the reasoning that females may lack the neck musculature and strength necessary to adequately support the neck in the high-G environment with HMDs. The risk of neck injury may be increased for these pilots. No subjects reported any injuries and most had only minor soreness early in testing. The females worked very hard, much harder than the males in this research sample, in order to perform comparably to the males; therefore, they might be expected to fatigue more quickly. They might be working at the edge of their physical capability, pressing the envelope to be successful. Any advantages they could gain from strength training of the neck muscles should be encouraged.

CONCLUSIONS

Electromyography used to obtain indications of neck muscle activity and its involvement in stabilizing the neck during $+G_z$ acceleration was successful in demonstrating differences related to a variety of factors including gender. Females in this particular research population do use more of their maximal capability than males when completing challenging exposures to simulated air combat maneuvers with increased helmet loads. The hypothesis cannot be rejected based on this result. It appears that females entering into the realm of high-performance aircraft and HMDs may well be at increased risk for neck injuries.

Any determination of cause for the gender differences is limited by the sample available. It appears that size may not be the primary factor because, in both males and females, smaller subjects were shown to be able to do better than some larger subjects. Anthropometry showing comparative neck sizes from which muscle cross-sectional area might be determined was not available.

One somewhat surprising result from this research was the apparent, relatively minor role of helmet weight and center of gravity at high G. It is obvious that the $+G_z$ acceleration and the resultant inertial forces are the primary influences acting against neck stability. This was very noticeable early in testing when subjects could not continue tracking in some of the higher G exposures. With $+G_z$ loading providing such an overpowering influence, it might be reasonable to expect that anything done to reduce weight and CG effect would be advantageous. Consideration should be given to the fact that USAF operations are performed at G levels that sometimes exceed 6.5 G, up to 9 G.

As long as pilots are able to maintain their heads in a neutral position while pulling the high Gs, the neck may be able to withstand inertial forces, but spinal compressive forces may still be problematic - acting on the cervical vertebrae and the intervertebral disks. It seems head movement in the cockpit should be limited if possible. Any requirement for head movement to operate new head or eye-slaved weapons systems may face difficulties and compromise pilot performance.

In spite of these findings, however, there is some reason to be optimistic. Performance of tracking tasks was not significantly affected. It remains to be seen if the tendency for females to have higher tracking error might be found significant in different populations or with more demanding tasks.

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