NECK MUSCLE ACTIVATION LEVELS DURING FRONTAL IMPACTS

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ABSTRACT

Helmet-mounted systems, such as night vision goggles and helmet-mounted displays, are designed to enhance pilot performance; however, they may also affect pilot safety during ejection due to the change in helmet inertial properties. The effects of variable helmet weight and bracing ability on subject response during impact are unknown. Electromyogram (EMG) is a useful tool to investigate the mechanics of bracing and the relationships to helmet weight and impact acceleration. In addition, EMG could be used to help establish the relationship between the potential for neck injury and the force exerted by the neck muscles due to bracing. EMG can be used to determine the activation timing of the muscles and to estimate the force produced by the muscles in a dynamic environment. A series of tests were conducted on a Horizontal Impulse Accelerator using male and female volunteers to investigate the effects of helmet weight on human response to short-duration frontal impacts of variable magnitude. Helmet weights ranged from 0 lb (no helmet) to 4.5 lbs, and acceleration levels were 6, 7, 8 and 10 g. The MyoMonitor Portable EMG System by DelSys was used to collect data from ten subjects. The electrodes were placed on the right and left upper trapezius and sternocleidomastoid. Amplitude and frequency components of the signals were evaluated to determine the amount of force exerted by the muscle. Root Mean Squared (RMS) amplitude analysis indicated that, in general, the muscular strain increased with increasing –Gx acceleration levels. The trapezius produced more force than the sternocleidomastoid. Activity of both muscle groups was synchronized, by their RMS values, with head and neck motion. A method of collecting EMG data during short-duration impact accelerations was developed. It was demonstrated that in fact the neck muscles can respond quickly to the short-duration impacts. Furthermore, the EMG system was able to collect these changes during the impact. Because of these facts and this unique research, further studies are warranted to establish a relationship between potential for neck injury and muscle force.

BACKGROUND

Tests by Buhrman and Perry at the Air Force Research Lab's Acceleration Effects and Escape Branch have evaluated the effects of variable helmet inertial properties on the biodynamic response of human volunteers exposed to vertical (+Gz) and lateral (+Gy) impact accelerations.[3,4] The objective of this study was to provide additional human dynamic data from a -Gx impact environment with a variable weight helmet. This is required to complete the development of multi-axial cervical injury criteria for the three coordinate axes, to continue the development of head/neck biodynamic models, and to continue the development of the

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biodynamic response database. The results of this program will contribute to the development of design guidelines for safe use of helmet systems that include devices that increase the weight and distribution of head-supported mass. The effects of subject bracing on human neck response have not been defined in the past, but it has been observed that the active neck musculature plays a significant role in reducing head/neck motion during impact. If we better understood the relationship between bracing and potential of injury we could provide detailed instructions to pilots during training so that in the event of an ejection they could lower their chances of injury with their position and brace.

Many different methods are utilized to determine the safety of a crewmember in dynamic environments. These include human volunteer testing, manikin testing, human surrogate testing, and modeling and simulation. Each has its own strengths and weaknesses. Recently, modeling and simulation have become viable methods to investigate the safety potential of different equipment and different environments without putting actual people at risk. Previously, due to computational limitations, the majority of human models was passive in nature and modeled only the involuntary response. This method proves quite reasonable when you only need to know the kinematics of the occupant or the overall global force response. However, when you want to investigate deeper into the types of injuries, a more realistic model is needed. This model will incorporate the active muscle response of the particular crewmember. This muscle response will include any voluntary bracing as well as a muscle activation that occurs as a result of the impact-induced stress. It should be noted that, although the computational models can simulate the active muscle response of the human, little data exist to validate these models so that they can be applied to real world problems. Experimental EMG data are necessary to provide accurate muscle activation signals for different impact simulations.[7]

The EMG can be collected during bracing and impact of human volunteers who are subjected to a variety of dynamic events. These data can then form the baseline from which several groups of muscles can be incorporated into previously passive models. EMG is also a useful tool to investigate the mechanics of bracing and the relationships to helmet weight and impact acceleration. In addition, EMG can be used to help establish the relationship between the potential for neck injury and the force exerted by the neck muscles due to bracing. EMG can be used to determine the activation timing of the muscles and to estimate the force produced by the muscles in a dynamic environment.

The role of upper trapezius and sternocleidomastoid (SCM) during long-duration head and neck loading situations has been studied. Tests by Butler found that the posterior muscles of the neck showed bursts of activity during whole-body vibrations that were synchronized with neck flexion.[2] The anterior muscles, however, showed little activity and no correlation with neck motion. Phillips and Petrofsky examined the characteristic changes in the EMG data from the upper trapezius and SCM associated with isometric muscle fatigue.[5,6] Past research does not quantify timing and level of muscular activation during short-duration impact events. The objective of this EMG pilot study was to develop a reliable method of collecting meaningful EMG data during short-duration impacts. The muscles of interest are those that play a major role in neck position and stability: the upper trapezius and SCM. A secondary objective was to collect preliminary EMG data on a subset of the –Gx subjects and use the information to determine activation timing of the muscle. Also desired were correlations between neck muscle force and impact level, and muscle force and helmet weight.

METHODS

A series of short-duration, frontal impact tests were completed at Wright Patterson AFB using the Horizontal Impulse Accelerator (HIA). The HIA has a 240-foot-long track and a 24-inch-diameter pneumatic actuator, and operates on the principle of differential gas pressures acting on both surfaces of a thrust piston in a closed cylinder. The impact acceleration occurs at the beginning of the experiment as stored high-pressure air is allowed to impinge on the surface of the thrust piston, thus propelling the sled. As the sled breaks contact with the thrust piston, the sled coasts to a stop or is stopped with a pneumatic brake system.

Male and female subjects were tested with approval obtained from the Wright Site Institutional Review Board. The HIA generated an acceleration impulse that approximated a half-sine wave with a pulse duration of 150 ms. Peak sled acceleration levels of 6, 7, 8 and 10 g were generated with total head-supported weight ranging from 0 (no helmet) to 4.5 lbs. The seat back and pan of the sled were not reclined. A PCU-15 or -16/P harness and HBU lab belt were used to restrain the subjects, all preloaded to 20 ± 5 lbs. A photo of the HIA facility (also referred to as the "sled track") is shown in Figure 1.

Figure 1. AFRL Horizontal Impulse Accelerator with Subject

The MyoMonitor Portable EMG System by DelSys was used to collect data from ten subjects. These data were collected separately from the rest of the response measurements, which included but were not limited to: head acceleration, chest acceleration, belt loads and seat loads. The sensors were placed on the right and left upper trapezius and sternocleidomastoid (SCM) over the belly of the muscles. A conductive gel was applied to the sensors which were then fixed to the subjects using a double-stick adhesive and covered with medical tape. A reference sensor was placed on the first thoracic vertebrae.

The EMG data acquisition was triggered at 6 seconds before impact by the master control station of the HIA and sampled at 1024 samples/s for 10 seconds. This allowed for synchronization of the EMG data with the impact event. At 2 seconds before impact, the subjects were instructed to brace with approximately 30 pounds of force in the rearward direction, with their head firmly against the headrest. A load cell was placed in the headrest to record the level of the bracing.

Amplitude and frequency components of the signals were evaluated to determine the amount of force exerted by the muscle.

RESULTS

EMG data (Figure 2) were collected for 29 tests with a total of 77 measurements made for both muscle groups of interest. Root Mean Squared (RMS) time histories (Figure 3) of the EMG voltage were calculated for each successful data collection. Non-zero resting voltages, also seen in Figures 2 and 3, were collected during the period before bracing and impact. These resting values would later be used for calculating normalization values that would be applied as a DC offset. The non-zero initial values represent the baseline activity in the muscles to keep the head supported as well as any normal signal activity in the electronics. No muscle activity was recorded in the SCMs during bracing, while measurable muscle activation occurred in the trapezius. This was to be expected since the bracing required of the subjects is that of an isometric extension action and the SCMs primarily act in flexion.

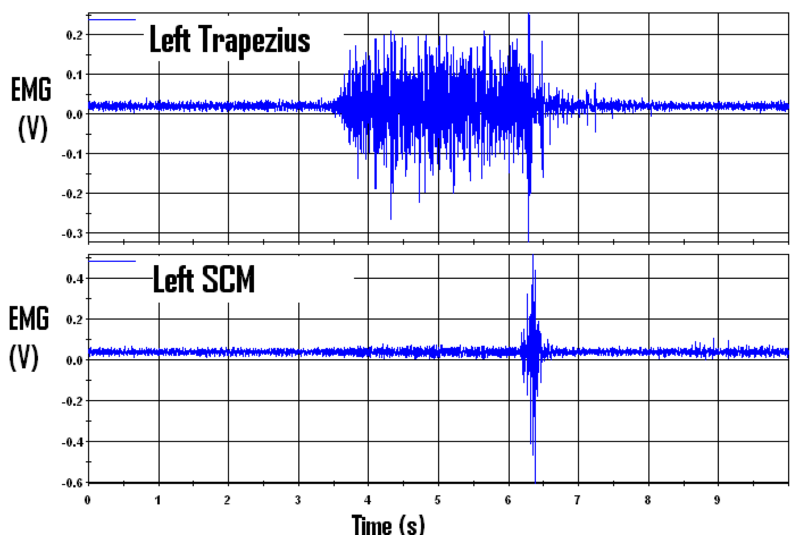

Figure 2. Example of Raw EMG data.

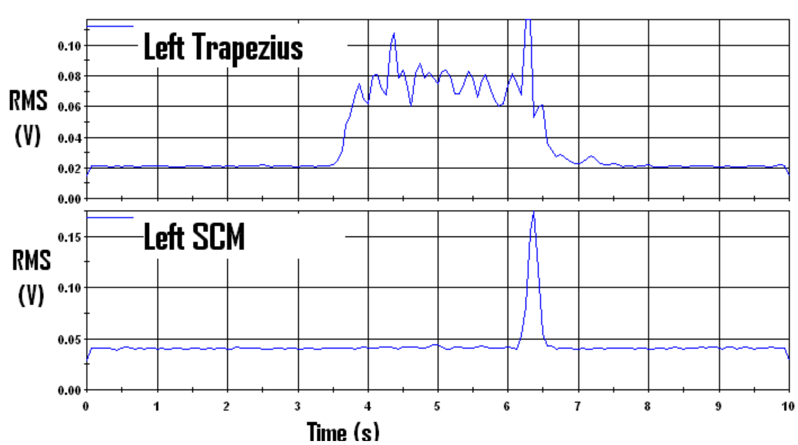

Figure 3. Example of Processed EMG data: Root Mean Squared.

To facilitate data processing, an automatic algorithm was developed that would detect the start and end of the bracing period and the start and end of the impact period. This was accomplished by use of visual data interpretation. The end of the bracing period was found by going in reverse time from the maximum voltage, which is due to impact, until it was found to have a negative slope between two data points after taking the root mean squared (RMS). The beginning of the bracing period is dependant upon each individual test and was therefore found using a difference tolerance that was initially estimated then graphed for user approval. The SCMs were not activated during bracing effect, so this phase was processed only for the trapezius. When the sensors were sufficiently affixed to the surface over the muscles being measured, the impact period was visibly defined. However, in an effort to avoid any residual effects of the impact period upon the post-impact calculations, the start time for post-impact begins at time of maximum voltage plus 0.625 seconds and extends the remaining collection time period. This was found to be clear of residual impact effects. An example of the processed EMG data with event marker can be found in Figure 4.

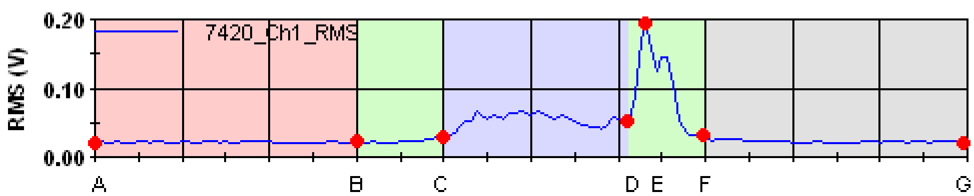

Figure 4: Example Processed EMG data with event markers

A – B: Resting period during countdown
C – D: Bracing period before impact
E: Time of Maximum Voltage
F – G: Resting period after impact

For the 6 and 8 g tests, the time of the peak muscle activation occurred at approximately the same time for both muscle groups; however, when the acceleration level was increased to 10g's, the trapezius was activated much sooner (Figure 5). One explanation for this is that at 10g's more head rotation occurs. This increased amount of head rotation brings the SCMs more into play and their activation becomes important in trying to restrain the head from further rotation. However, this later activation timing did not depend on the amount of bracing activation present in the trapezius, meaning that the timing of activation was more a consequence of the stimulus and not of the initial conditions (Figure 6). There was an increase in activation timing from bracing level when comparing the 6g test to the 10g test; however, at 8g's the opposite was true.

Muscle activation levels were determined by scaling the bracing voltage to a constant force of 40 lbs. Peak activation levels were then calculated as the increase over bracing levels. Because this calculation involved the bracing levels, only the activity of the trapezius was calculated. Testing with a constant weighted helmet of 4.5 lbs showed a decrease in muscle activation from 6g's to 7g's, then an increase in muscle activation to 8g's (Figure 7). While it would be expected for the muscle activation to continue to increase with g level, it should be noted that the subjects are tested in a predefined order such that they start with the less severe test first, then work their way up to the tests at higher g level and heavier helmet. In fact, the average headrest load measured during bracing increased with increasing impact levels while wearing the 3.0 lb helmet (the first testing that occurred in sequence). This lower level of bracing would tend to increase the peak activation ratio. A further point is that the 6 G data with the 4.5 lb helmet is that from a single subject. The neck muscle activation level generally increased when heavier helmets were worn (Figure 8). This was to be expected since the bracing levels were similar and the heavier helmet would require higher activation levels to keep the head stable and prevent head rotation during impact.

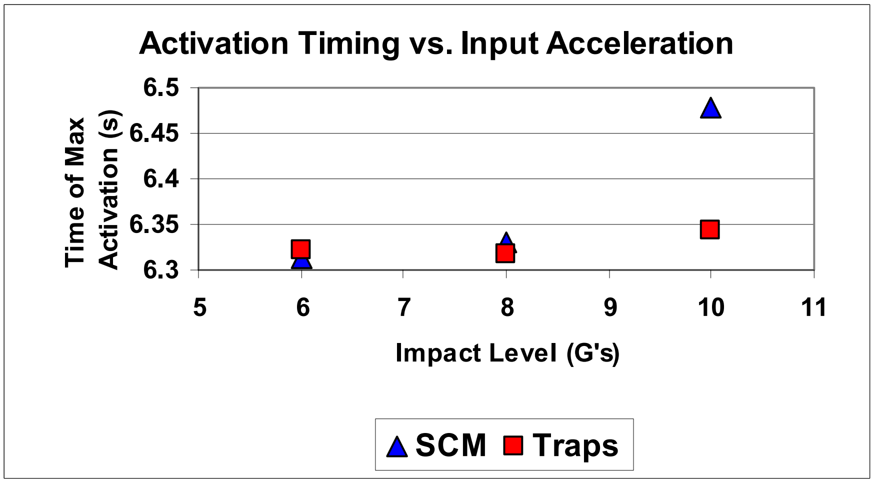

Figure 5. Neck muscle activation timing vs. input acceleration.

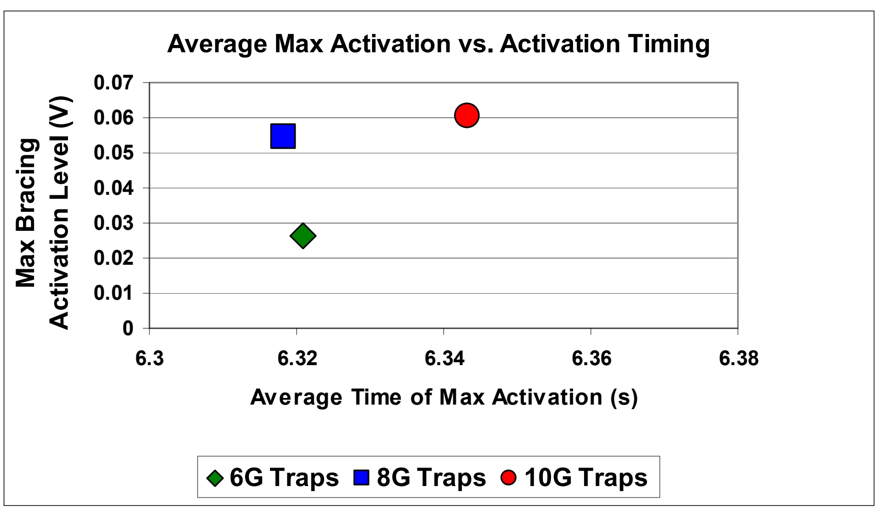

Figure 6. Activation level vs. activation timing

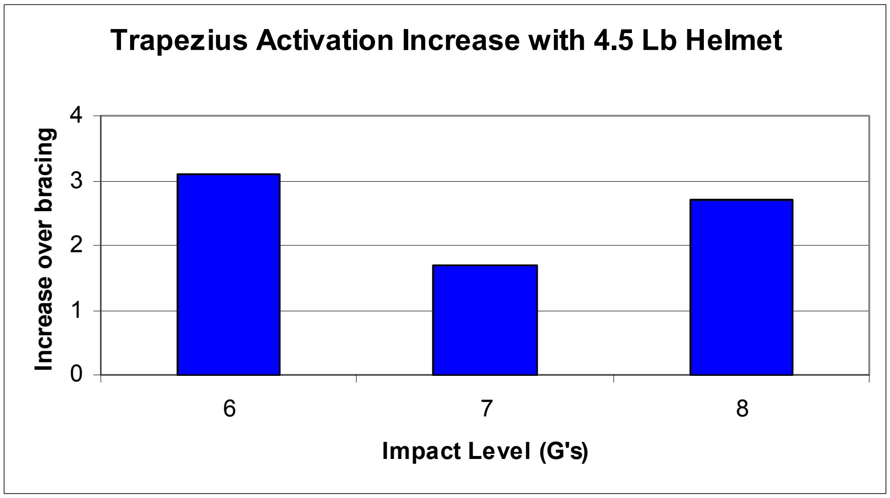

Figure 7. Change in muscle activity with change in impact level.

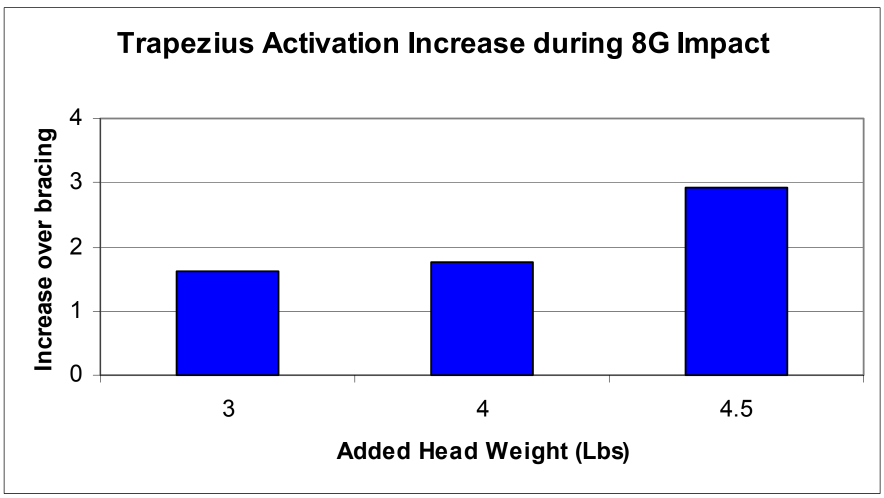

Figure 8. Increase of muscle activity level with change in helmet weight

DISCUSSION

One interesting finding in this program was that as the subjects were exposed to increasing accelerations, their bracing force also increased. The subjects were trained to brace to a predefined level and were instructed during each test to brace to this level; however, the subjects were aware of the impending impact level and motivational factors played a role in how forcibly they braced. This finding also holds true for the full test series with the subjects whose muscle activation level was not measured (Figures 9 and 10).[1] It should be noted that the testing sequence was not just that of an increasing g level, but also of increasing helmet weight. One area that made the data difficult to interpret was the fact that even with this increasing bracing force, no significant change in the muscle activation level was detected. It was because of this that different methods of normalizing the data to account for the different levels of voluntary effort exerted were explored. It would have been beneficial to record Maximum Voluntary Contractions (MVC) on the subjects just prior to impact and measure the exerted force.

However, there were some initial concerns over the safety of the subjects and the possibility that they had fatigued neck muscles before they were exposed to impact accelerations.

A method of collecting neck muscle activity data from the trapezius and SCM during short-duration impact experiments was successfully developed. Amplitude analysis of the preliminary data collected indicated that, in general, the muscular strain increased with increasing –Gx acceleration levels. The trapezius produced more force than the sternocleidomastoid. Activity of both muscle groups was synchronized, by their RMS values, with head and neck motion. This study showed that the muscles in the neck can and do react during the short time of these frontal impacts. Now that this information is known, future studies can concentrate on further characterizing the roles of the different muscle groups to the biomechanic response of the head and neck system.

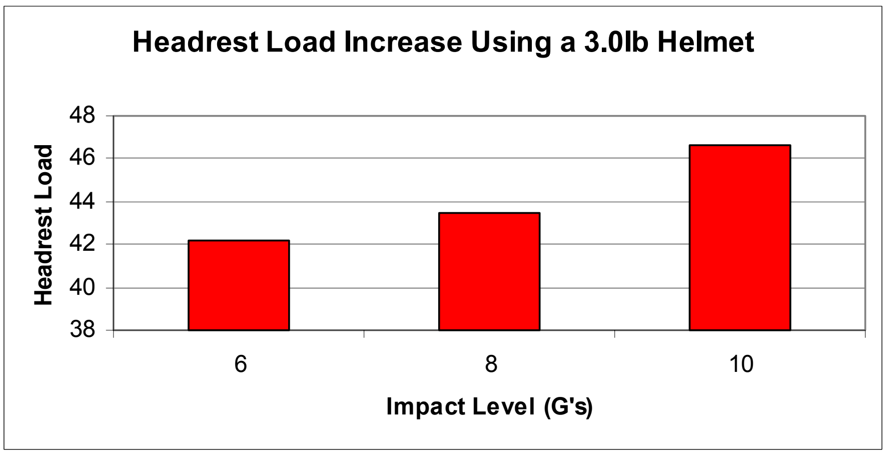

Figure 9. Increase of Headrest Loads across All Subjects with Change in Impact Level Using a 3.0 lb Helmet

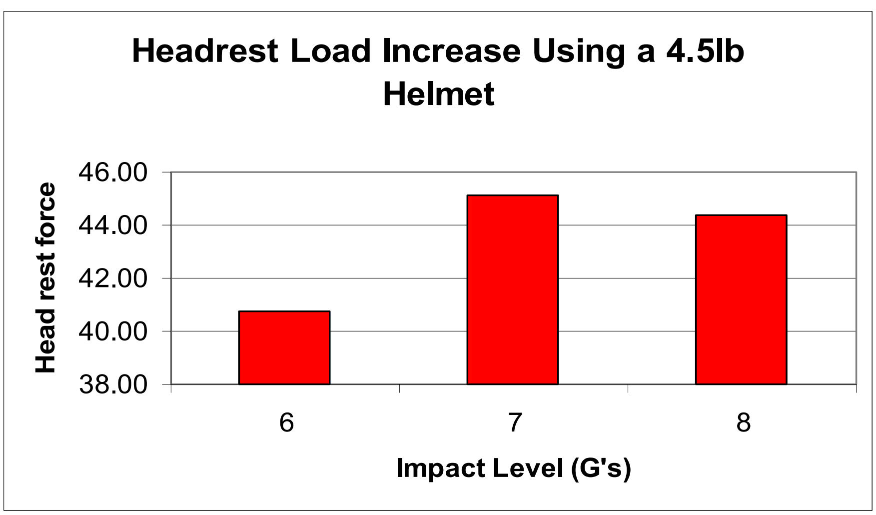

Figure 10. Increase of Headrest Loads across All Subjects with Change in Impact Level Using a 4.5 lb Helmet

REFERENCES

1. Buhrman J.R., Doczy E.J. and Mosher S.E. The Effects of Variable Helmet Weight on Head Response and Neck Loading During Frontal –Gx Impact. Proceedings of the 42nd SAFE Symposium. 2004.
2. Butler B.P. Helmeted Head and Neck Dynamics Under Whole-Body Vibrations. Doctor of Philosophy dissertation. The University of Michigan. 1992.
3. Perry C.E. and Buhrman J.R. Effect of Helmet Inertial Properties of the Biodynamics of the Head and Neck During +Gz Impact Accelerations. SAFE Journal, 26(2): 34-41, 1996.
4. Perry C.E., Buhrman J.R., Doczy E.J. and Mosher S.E. Evaluation of the Effects of Variable Helmet Weight on Human Response During Lateral +Gy Impact. AFRL-HE-TR-2004-0013, August 2003.
5. Phillips C.A. and Petrofsky J.S. Neck Muscle Loading and Fatigue: Systematic Variation of Headgear Weight and Center-of-Gravity. Aviation, Space, and Environmental Medicine. October 1983.
6. Phillips C.A. and Petrofsky J.S. Quantitative Electromyography: Response of the Neck Muscles to Conventional Helmet Loading. Aviation, Space, and Environmental Medicine. May 1983.
7. van der Horst M.J., Thunnissen J.G.M., Happee R., van Haaster R.M.H.P. and Wismans J.S.H.M. The Influence of Muscle Activity on Head-Neck Response During Impact. Society of Automotive Engineers, Inc. 1997.

BIOGRAPHIES

Joseph Pellettiere is a mechanical engineer for the Biomechanics Branch, Human Effectiveness Directorate, Air Force Research Laboratory. He has a BS in Biomedical Engineering and an MS in Mechanical Engineering from Case Western Reserve University, and a Ph.D. in Mechanical Engineering from the University of Virginia. His experience is in biomechanics, human simulation and injury, crash protection and prevention using both testing and computational technologies. He currently leads the modeling simulation group in the branch.

Erica Doczy is a biomedical engineer for General Dynamics supporting the Biomechanics Branch, Human Effectiveness Directorate, Air Force Research Laboratory. She has a BS in biomedical engineering from Wright State University. Her experience is in impact biomechanics and human systems test and evaluation. She is currently the associate investigator of a study examining the effects of helmet weight during vertical impacts using manikin and human volunteer subjects.

Mary Ann Sanders is a Second Lieutenant in the United States Air Force. She is a member of the Air Force Research Laboratory, Human Effectiveness Directorate. She graduated and commissioned from Clemson University in August 2003, and has a BS in Mathematical Sciences with an option in Applied Analysis. Her training and experience are in the field of electromyography and mathematical programming.