Effects of Pre-Impact Bracing on Chest Compression of Human Occupants in Low-Speed Frontal Sled Tests

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ABSTRACT

Continued development of computational models and biofidelic anthropomorphic test devices (ATDs) necessitates further analysis of the effects of muscle activation on the biomechanical response of human occupants in automotive collisions. The purpose of this study was to investigate the effects of pre-impact bracing on human occupant chest compression during low-speed frontal sled tests. In this study, a total of 10 low-speed frontal sled tests (5.0g, Δv=9.7kph) were performed with 5 male human volunteers. The height and weight of the human volunteers were approximately that of the 50th percentile male. Each volunteer was exposed to 2 impulses, one relaxed and the other braced prior to the impulse. A 59 channel chestband, aligned at the nipple line, was used to measure anterior-posterior sternum deflection for all test subjects. Subject head accelerations, spine accelerations, and forces at each interface between the subject and test buck were recorded for all tests. A Vicon motion analysis system, consisting of 12 MX-T20 2 megapixel cameras, was used to quantify subject 3D kinematics (±1 mm) at a sampling rate of 1 kHz. The chestband data showed that bracing prior to the initiation of the sled pulse essentially eliminated thoracic compression due to belt loading for all subjects except one. The load cell data indicate that forces were distributed through the feet, seatpan, and steering column as opposed to the seatbelt for the bracing condition. In addition, the forward excursion of the elbows and shoulders were significantly reduced during the braced condition compared to the relaxed condition. The data from this study illustrates that muscle activation has a significant effect on the biomechanical response of human occupants in frontal impacts and can be used to refine and validate computational models and ATDs used to assess injury risk in automotive collisions.

INTRODUCTION

Nearly 30,000 passenger vehicle occupant deaths occur annually in the United States. Approximately 50% of these fatalities are due to frontal crashes (NHTSA Traffic Safety Facts, 2008). In addition, the number of occupants sustaining injuries greatly exceeds the number of fatalities.

Computational models and anthropomorphic test devices (ATDs) are commonly used to predict and evaluate human occupant responses and injuries in motor vehicle collisions. These research tools are primarily validated against post mortem human surrogate (PMHS) data, which do not include the effects of muscle activation. However, studies have shown that tensing muscles prior to a crash event can change the kinetics and kinematics during the crash (Armstrong et al., 1968; Hendler et al., 1974; Begeman 1980; Sugiyama 2007; Ejima 2008). There have been no studies to the authors' knowledge that have quantified the chest deflection of human volunteers in a relaxed and braced state. Therefore, the purpose of this study was to investigate the effects of pre-impact bracing on human occupant chest compression during low-speed frontal sled tests.

METHODS

A total of 10 low-speed frontal sled tests (5.0g, Δv=9.7kph) were performed with 5 male human volunteers. Selected volunteers were approximately 50th percentile male height and weight (175cm; 76.7kg) (Schneider et al., 1983). Approval to conduct the human subject testing presented in the current study was granted by the Virginia Tech Internal Review Board (IRB). In addition, all volunteers signed an informed consent form prior to participating in the study.

Experimental Setup

Dynamic frontal sled tests were performed using a custom mini-sled and test buck (Figure 1). The mini-sled was accelerated with the use of a pneumatic piston, which was used to generate a 5.0 g (Δv=9.7kph) frontal sled pulse (Figure 2). The sled pulse severity was determined based on previous research which has shown that a frontal sled pulse of this severity does not result in injury (Arbogast et al., 2009).

The test buck was instrumented with 5 multi-axis load cells and 18 single-axis accelerometers. A six-axis load cell was installed on both the seatpan and seatback (Robert A. Denton, Inc., 44 kN-Model 2513, Rochester Hills, MI). A six-axis load cell was installed on the right foot support (Robert A. Denton, Inc., 13.3 kN-Model 1794A, Rochester Hills, MI) and left foot support (Robert A. Denton, Inc., 13.3 kN-Model 1716A, Rochester Hills, MI). A five-axis load cell was installed on the steering column (Robert A. Denton, Inc., 22.2 kN-Model 1968, Rochester Hills, MI). Three-axis accelerometer cubes (Endevco 7264B, 2000 g, San Juan Capistrano, CA) were rigidly mounted to each load cell plate for inertial compensation. The test buck acceleration was measured with a three-axis accelerometer cube (Endevco 7264B, 2000 g, San Juan Capistrano, CA) rigidly mounted to the frame under the seatpan. Belt load sensors were added to the retractor, shoulder, and lap belts (Robert A. Denton, Inc., 13.3 kN-Model 3255, Rochester Hills, MI). Seatbelt spool out at the retractor was measured with a potentiometer (Space Age Control Inc. 160-1705, 539.75mm, Palmdale, CA) attached to a custom seatbelt clamp.

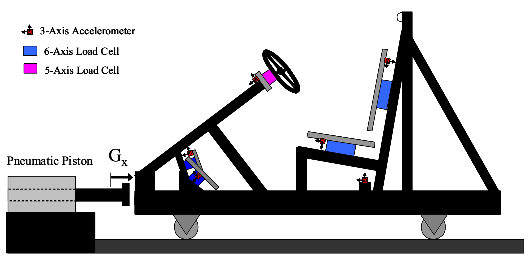

Figure 1: Test buck schematic.

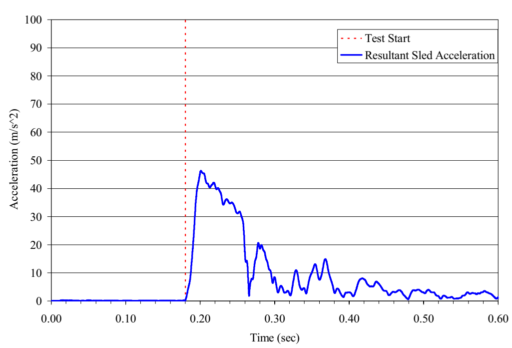

Figure 2: Resultant sled acceleration.

Subject Instrumentation

A 59 channel chestband (Denton, Inc., Model 8641, Rochester Hills, MI) was used to quantify thoracic displacement and contours. Gage 1 was positioned at the spine and the band was wrapped around the chest so that gage number ascended in the clockwise direction. The chestband was aligned with the nipple line (Figure 3). Heavy duty double sided adhesive tape was used to securely attach the chestband to each subject. The chestband was subsequently wrapped with co-flex, a self-adherent compression wrap, to ensure a tight connection and protect the chestband wires throughout each trial. The closest gage to both the sternum and the spine were noted as well as their respective distances.

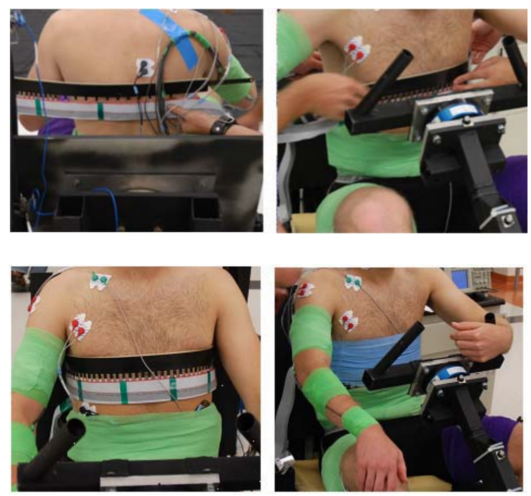

Figure 3: Chestband aligned with the nipple line of the chest.

For each subject, accelerations were measured at the sternum, spine, and head. In order to obtain linear acceleration and angular velocity of the head, three orthogonal accelerometers (Endevco 7264B, 2000 g, San Juan Capistrano, CA) and a three-axis angular rate sensor (IES 3103, 4800 deg/s, Braunschweig, Germany) were mounted to a metal mouthpiece fixture. A “boil and bite” mouth guard of thermoplastic material was created for each subject to ensure proper, repeatable positioning as well as a tight junction with the subject. In addition, each subject wore a chin strap to ensure that the jaw remained tightly clamped on the mouthpiece during the trials. Three-axis accelerometer cubes (Endevco 7264B, 2000 g, San Juan Capistrano, CA) were attached to the sternum (at the suprasternal notch), the 7th cervical vertebrae (C7), and the sacrum of each subject. The three-axis accelerometer cubes were individually attached to the skin with adhesive patches glued to the cubes. However, since the focus of the current study is chest compression, the results of subject acceleration data are not presented.

Surface electromyography (EMG) was obtained from a total of 20 muscles during each test event (Table 1). Several preparation steps were taken prior to attaching the electrodes to the skin. To ensure a strong bond with the skin, hair was removed at the location of electrode attachment. The skin was lightly abraded to remove dead epithelial cells and then wiped with isopropyl alcohol to remove oils and surface residue. The skin was allowed to dry before the electrodes were adhered. After connecting the wires to the electrodes, the electrodes were wrapped with co-flex, a self-adherent compression wrap, to ensure a tight connection. However, since the focus of the current study is chest compression, the results of EMG data are not presented.

Table 1: Target muscles for EMG.

Body Region	EMG Location
Arm	Right Biceps Brachii
	Right Triceps Brachii
Forearm	Right Flexor Carpi Radialis
	Right Extensor Digitorum
Neck & Shoulder	Right Sternoceidomastoid
	Right Trapezius
	Right Deltoid
Chest	Right Pectoralis Major
Abdomen	Right Rectus Abdominis
	Right External Abdominal Oblique
Spine	Right Upper Erector Spinae
	Right Lower Erector Spinae
Thigh	Right/Left Rectus Femoris
	Right/Left Biceps Femoris
Shank	Right/Left Tibialis Anterior
	Right/Left Gastrocnemius

Test Conditions

Each volunteer was exposed to two 5.0 g (Δv=9.7kph) frontal sled impulses, one relaxed and the other braced prior to the impulse. For all tests, a load limiting driver side seatbelt was placed around the test subject, and the slack was removed. Volunteers were informed before each test as to whether they were to remain relaxed or brace themselves for the sled impulse. For the relaxed tests, a television monitor was used as a distraction mechanism and the trigger was out of sight so that the volunteers were unaware of when the test would occur. Prior to the relaxed tests, subjects were instructed to relax and continue to watch the monitor while facing forward. Then, the sled pulse was randomly initiated after several minutes of quiet sitting. For the tensed tests, subjects were asked to brace themselves with both their arms and legs. A guided countdown was used to instruct the volunteers when to brace with their arms and legs prior to the initiation of the sled pulse. Each subject had a waiting time of approximately 30 minutes between subsequent test conditions.

Data Acquisition and Processing

An onboard data acquisition system was used to record 148 channels of data at a sampling rate of 20kHz. For relaxed tests, data was collected during the relaxed state and test event. For braced tests, data was collected during the relaxed state, braced state, and test event. Data included subject accelerations, test buck accelerations, chest contour, surface electromyography of 20 muscles (legs, arms, abdomen, back, and neck), and forces at each interface between the subject and test buck. All reaction load cell data were compensated for crosstalk. The forces measured by each reaction load cell were inertially compensated with the use of three single-axis accelerometers mounted to each reaction plate. The reaction load cell data and test buck accelerometer data were filtered using SAE Channel Filter Class (CFC) 60 [SAE J211, 1995]. The seatbelt load cell data were filtered using SAE Channel Filter Class (CFC) 180. The unfiltered chestband data was processed using a software package called RBandPC. This software was used to generate 2D chest contours in 10 ms increments as well as the deflection between the spine and sternum.

A Vicon motion analysis system, consisting of 12 MX-T20 2 megapixel cameras, was used to quantify the 3D kinematics ($\pm$1mm) of photo-reflective markers placed on both the test subject and the test buck at a sampling rate of 1kHz (Figure 4). Marker trajectories were converted to the reference frame of the test buck and then to the SAE J211 sign convention [SAE J211, 1995]. The displacements of selected anatomical regions of interest relative to the buck were categorized as global trajectories. Regions of interest included the upper extremities (elbows and shoulders), lower extremities (knees), pelvis (hips), and the head. The excursions of these regions were normalized to their respective initial positions and compared by test condition across subjects. A paired Student's t-test was used to assess significance between relaxed and braced conditions for each volunteer.

High-speed video was obtained from the lateral side of the volunteer at a sampling rate of 1,000 Hz with the use of a high resolution, high light sensitivity camera (Vision Research, Phantom V-9, Wayne, NJ).

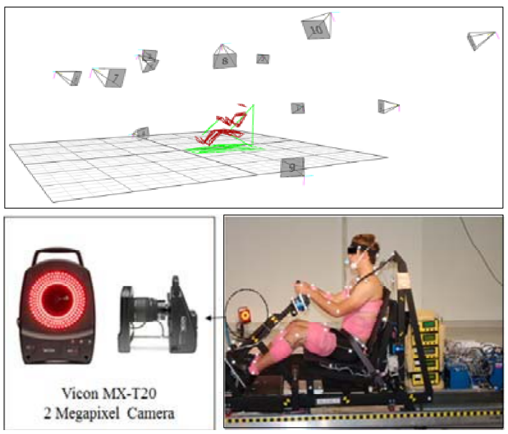

Figure 4: Vicon System and markers.

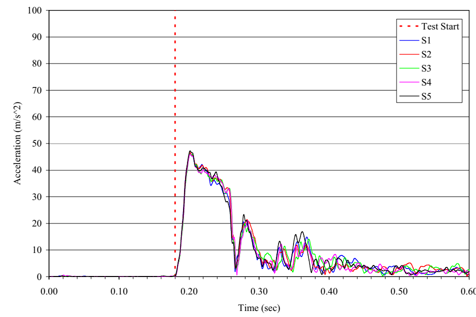

Figure 5: Resultant sled accelerations for relaxed tests.

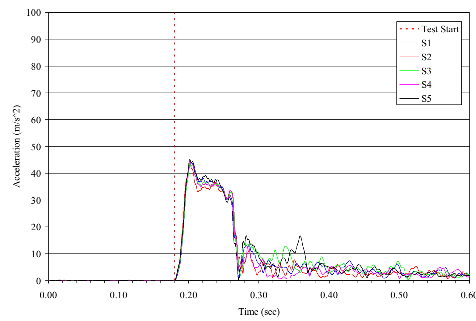

Figure 6: Resultant sled accelerations for braced tests.

RESULTS

The time histories of the resultant sled accelerations were calculated and plotted for each subject and test condition (Figures 5 and 6). The consistency in the resultant sled acceleration between subjects for a given test condition shows that the custom mini-sled was extremely repeatable. Exemplar high-speed video stills for each condition are provided for qualitative kinematics (Figure 7). The sled pulse started at t=180ms for all tests.

Qualitative examination of the global trajectories of each test subject revealed marked differences between the relaxed and braced conditions in both initial position and forward excursions of the volunteer occupants (Figure 8). The normalized data highlighted pronounced differences in forward excursions due to bracing. It was found that bracing significantly reduced the forward excursion of the elbows ($p<0.01$), shoulders ($p<0.01$), and head ($p<0.01$). Although not significant, bracing considerably reduced the forward excursion of the lower extremities and pelvis.

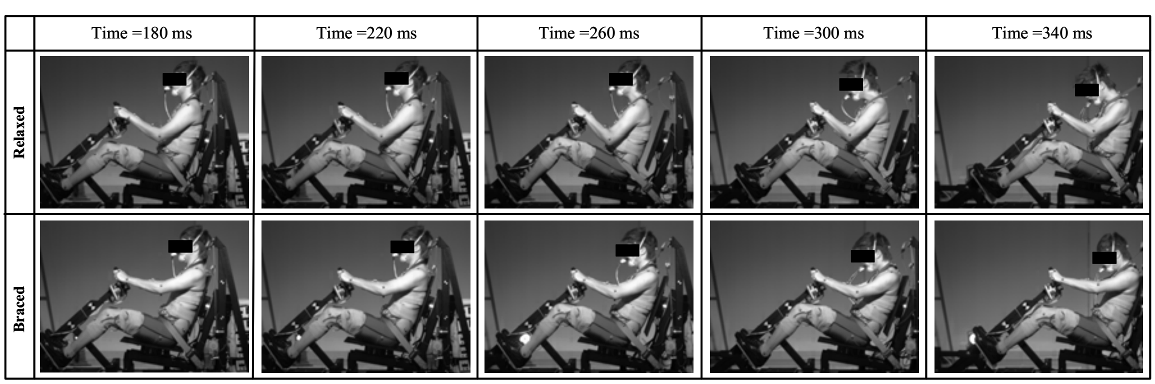

Figure 7: Exemplar high-speed video stills for relaxed and braced conditions.

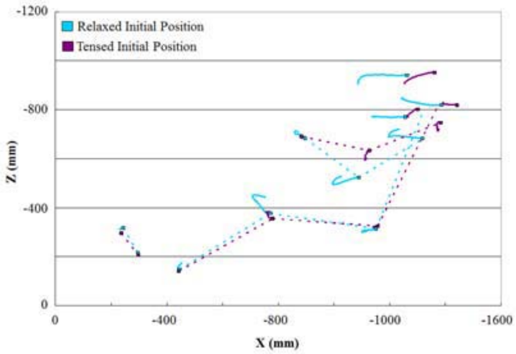

Figure 8: Example global trajectory of a relaxed vs. braced volunteer test.

Example 2D chestband contours were plotted for each test condition (Figures 9 and 10). In each figure, the red line represents the subject's chest contour 10 ms before the start of the trial and the blue line represents the subject's chest contour at the time of peak sternum deflection. For the braced trial, the red line represents the subject's chest contour while in a braced state, while the green line represents the subject's chest contour prior to bracing, i.e. relaxed state.

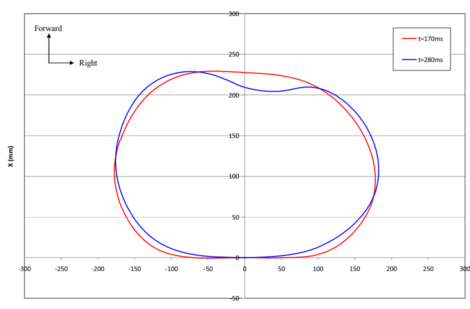

Figure 9: Example chestband contours during relaxed condition.

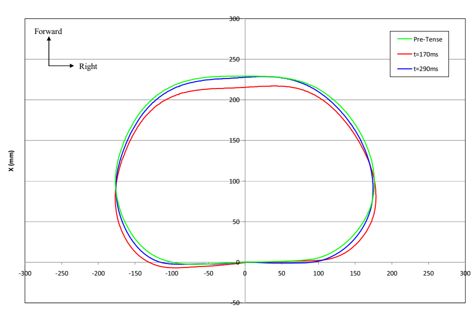

Figure 10: Example chestband contours during braced condition.

The chest compression for each subject was plotted for each test condition (Figures 11 and 12). Chest compression was defined as the ratio of the instantaneous chest depth at the sternum to the chest depth at the sternum measured during the relaxed state prior to the test. Positive chest compression indicates a decrease in chest depth at the sternum, while negative chest compression indicates an increase in chest depth at the sternum.

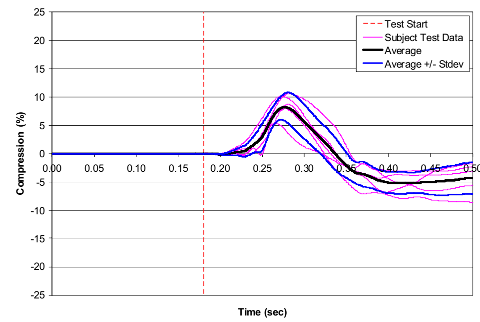

Figure 11: Chest compression during relaxed condition.

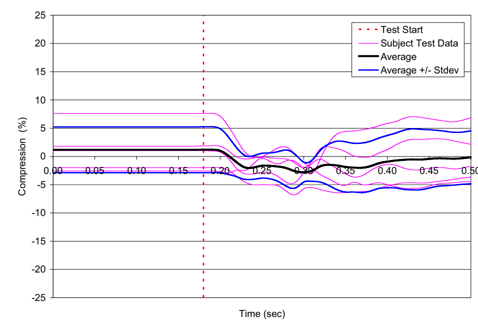

Figure 12: Chest compression during braced condition.

DISCUSSION

The chestband data show that the chest was compressed due to thoracic belt loading for all subjects during the relaxed condition. For the braced condition, the chestband data show that the act of bracing prior to the test resulted in a reduction in sternum depth, relative to the relaxed state, for three subjects due to compression against the seatback. For two of the subjects, the act of bracing prior to the test resulted in an increase in sternum depth, relative to the relaxed state, due to either lateral thoracic compression from the upper extremities or expansion of the pectoralis major muscles. During the test event for the braced condition, the chest depth at the sternum returned to a depth close to that recorded during the relaxed state for two of the subjects due to the decreased force applied to the seatback during the frontal sled pulse. For two subjects the chest depth at the sternum increased slightly due to lateral thoracic compression from the upper extremities. Regardless, chest compression due to thoracic belt loading was essentially eliminated during the braced condition for all but one subject. The reaction load cell data showed that the one subject who sustained minor chest compression due to belt loading (1.7% compression), exerted considerably lower bracing forces on the steering column and foot rests than all other subjects.

SUMMARY

In the current study, a total of 10 low-speed frontal sled tests (5.0g, Δv=9.7kph) were performed with 5 male human volunteers. The height and weight of the human volunteers were approximately that of the 50th percentile male. Each volunteer was exposed to 2 impulses, one relaxed and the other braced prior to the impulse. A 59 channel chestband, aligned at the nipple line, was used to measure anterior-posterior sternum deflection for all test subjects. The chestband data showed that bracing prior to the initiation of the sled pulse eliminated thoracic compression due to belt loading for all subjects except one. The load cell data indicate that forces were distributed through the feet, seatpan, and steering column as opposed to the seatbelt for the bracing condition. In addition, the forward excursion of the elbows and shoulders were significantly reduced during the braced condition compared to the relaxed condition. The data from this study illustrates that muscle activation has a significant effect on the biomechanical response of human occupants in frontal impacts and can be used to refine and validate computational models and ATDs used to assess injury risk in automotive collisions.

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