Effects of Muscle Activation on Occupant Kinematics in Frontal Impacts

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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 an occupant's biomechanical response in car crashes. In this study, a total of 28 dynamic sled tests were performed, 14 low (2.5g, Δv=4.8kph) and 14 medium severity (5.0g, Δv=9.7kph), with 5 male human volunteers (approximately 50th percentile male height and weight) and a Hybrid III 50th percentile male ATD. Each volunteer was exposed to 2 impulses at each severity, one relaxed and the other braced prior to the impulse. The ATD was subjected to 4 impulses at each severity. A 172 channel onboard data acquisition system was used to record subject head accelerations, spine 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 at a sampling rate of 20kHz. A Vicon motion analysis system, consisting of 12 MX-T20 2 megapixel cameras, was used to quantify subject 3D kinematics (±1mm) at a sampling rate of 1kHz. The excursions of select anatomical regions were normalized to their respective initial positions and compared by test condition across subjects. At the low severity, bracing significantly reduced ($p<0.05$) the forward excursion of the lower extremities (50%), pelvis (60%), upper extremities (35-70%), and head (47%). At the medium severity, bracing significantly reduced ($p<0.05$) the forward excursion of the upper extremities (63-69%) and the head (36%). Although not significant, bracing at the medium severity considerably reduced the forward excursion of the lower extremities (18-26%) and pelvis (22-25%). Data indicates that loads were distributed through the feet, seatpan, and steering column as opposed to the seatbelt for the bracing condition. This study illustrates that muscle activation has a significant impact on the biomechanical response of human occupants in front impacts.

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 2009). The number of occupants sustaining injuries greatly exceeds fatalities in magnitude. Frontal crashes present an important focus for research given the severity and global impact. Computational models and anthropomorphic test devices (ATDs) are commonly used to predict and evaluate human occupant responses in motor vehicle collisions. These research tools are currently validated against post mortem human subject (PMHS) studies. Therefore, these do not include the effects of muscle activation. Studies have shown that tensed muscles can change the kinematics and subsequently the kinetics during a crash (Begeman 1980; Sugiyama 2007; Ejima 2008). These findings infer that it is essential to investigate the effect of muscle activation on the biomechanical response of occupants in car crashes. Current studies that do investigate muscle activity are limited mostly to rear impacts, whiplash, and the application of limited or passive muscle forces (Brault 2000; Blouin 2003; Stemper 2004; Chang 2009; Siegmund and Blouin 2009). Within these studies, the number of targeted muscle groups is limited. Therefore, the purpose of this study was to investigate the effects of systemic dynamic muscle tension on the occupant kinematics and kinetics in low-speed frontal sled tests.

METHODS

A total of 28 dynamic frontal sled tests, 14 low (2.5g, Δv=4.8kph) and 14 medium severity (5.0g, Δv=9.7kph), were performed with 5 male human volunteers and a Hybrid III 50th percentile male ATD using a custom mini-sled and test buck accelerated by a pneumatic piston (Figure 1). Selected volunteers were approximately 50th percentile male height and weight. Each volunteer was exposed to 2 impulses at each severity, one relaxed and the other braced prior to the impulse. The ATD was subjected to 4 impulses at each severity. Subjects were positioned in the center of the test seat (right to left) with the feet centered on the foot plates and hands on the steering column. The average initial position of the volunteers in the relaxed trials was used to position the ATD on the test buck.

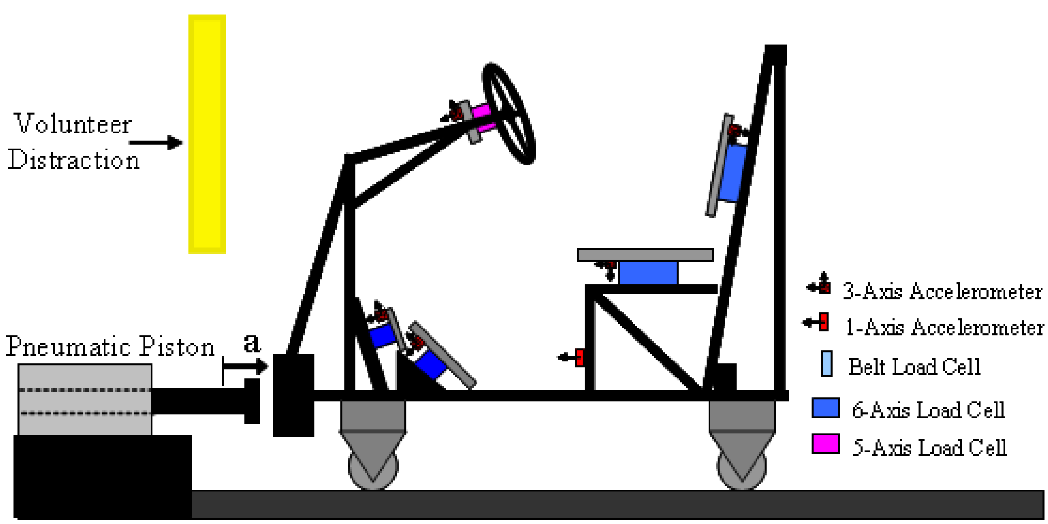

Figure 1: Test buck schematic with instrumentation.

Prior to each test, the human volunteers were informed 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 so that the volunteers were unaware of when the test would occur. For the braced tests, a countdown was used to instruct the volunteers when to brace with their arms and legs prior to the initiation of the sled pulse. A wait time of approximately 30 minutes between tests for an individual subject was employed.

A 172 channel onboard data acquisition system was used to record subject head accelerations, spine accelerations, chest contour, surface electromyography of 20 muscles (legs, arms, abdomen, back, and neck) (Figure 2), and forces at each interface between the subject and test buck at a sampling rate of 20kHz. High-speed video was captured from the subjects' lateral side at a sampling rate of 1kHz with the use of a high resolution, high light sensitivity camera (Vision Research, Phantom V-9, Wayne, NJ). Analysis of EMG, chest contour, acceleration and force data were not the focus of this manuscript.

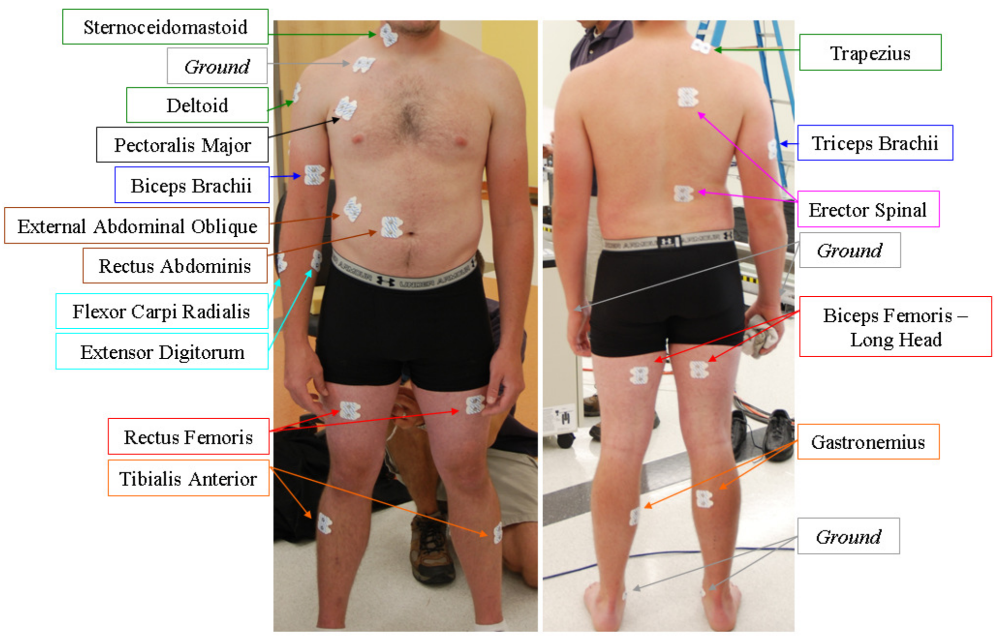

Figure 2 : Human volunteer EMG target muscles.

A Vicon motion analysis system, consisting of 12 MX-T20 2 megapixel cameras, was used to quantify the 3D kinematics (±1mm) of 43 photo-reflective markers placed on the test subject at key anatomical locations as well as the test buck at a sampling rate of 1kHz (Figure 3). A minimum of 2 cameras were required to triangulate the position of the markers. Marker trajectories were converted to the reference frame of the test buck, then to the SAE J211 sign convention (SAE, 1995).

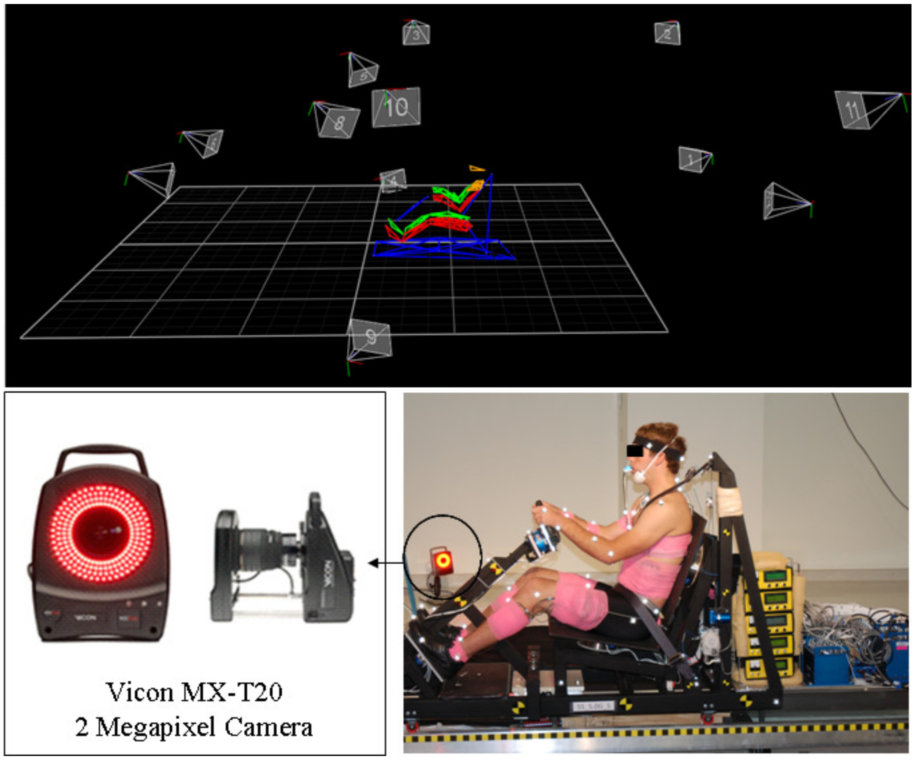

Figure 3: Experimental setup with subject on test buck.

After completing the four test configurations, additional anatomical markers were added and a quiet standing trial (100 Hz, 15,000 frames) was performed with the subject in the anatomical position with arms adducted and abducted for use in determining ankle, hip, and shoulder joint centers with a rigid body approach (Bell 1990; De Leva 1996). The knee joint center was determined by averaging the positions of the medial and lateral femoral epicondyles at each time step. Similarly, the elbow joint center was determined at each time step by averaging the positions of the medial and lateral humeral epicondyles. The wrist joint center was determined by averaging the radial and ulnar styloid process positions at each time step. The C7 location was determined by averaging the C7-Superior and C7-Inferior marker positions. Head CG was calculated throughout each trial using a rigid body approach and 2-D pre-test pictures.

Excursions of select anatomical regions were subsequently normalized to their respective initial positions. Regions of interest included the upper extremities (elbows and shoulders), lower extremities (knees), pelvis (hips), and the head (head CG). Peak forward (x) excursions of the select regions of interest were determined for each trial. All trajectory data was cut at the respective peak forward excursions. A paired Student's t-test was used to assess significance between relaxed and braced conditions at low and medium severity impulses for the human volunteers. A paired Student's t-test was also used to assess significance between low and medium severity impulses for each volunteer. Additional Student's t-tests for the mean, assuming unequal variances, were performed to assess significance between the ATD response and the volunteer relaxed and braced responses.

RESULTS

Global trajectory comparisons of volunteer relaxed and braced conditions were made qualitatively by examining sagittal (z vs. x) plane excursions (Figure 4). A change in initial position was observed between the two muscle conditions for all volunteers. All regional excursions for both volunteer muscle conditions and the ATD, with the exception of tensed right elbow, were significantly larger ($p<0.05$) for the medium severity impulse than the low severity impulse (Figure 5).

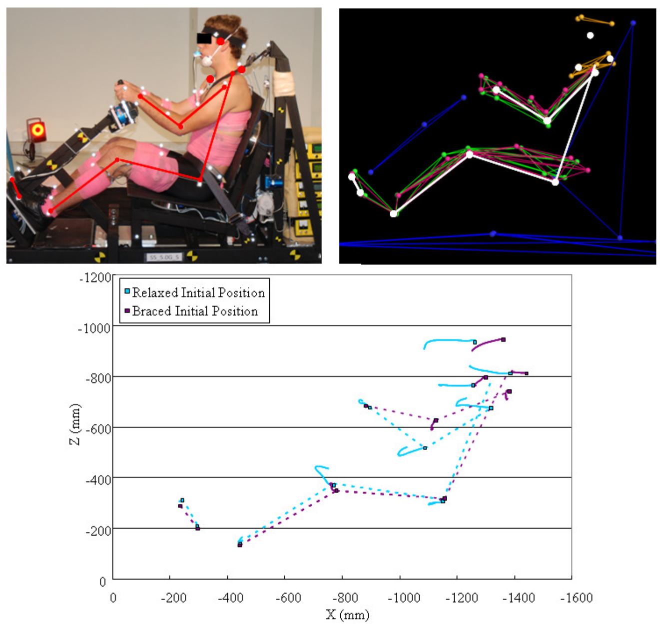

Figure 4: Representative comparison plot of relaxed and braced volunteer global trajectories.

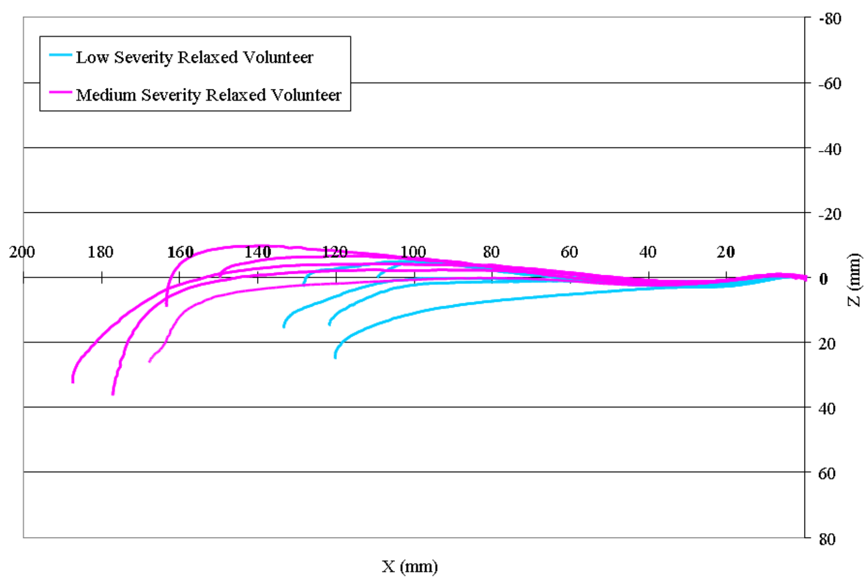

Figure 5: Representative comparison plot of low and medium severity volunteer excursions.
Note: Normalized head CG excursion shown.

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Muscle bracing elicited a significant effect on the forward excursions in the volunteer trials (Table 1). At the low severity, bracing significantly reduced ($p<0.05$) the forward excursion of the lower extremities, pelvis, upper extremities, and head. At the medium severity, bracing significantly reduced the forward excursion of the upper extremities and the head. Although not significant, bracing at the medium severity considerably reduced the forward excursion of the lower extremities and pelvis. A representative plot illustrating the decreased forward excursion as a result of bracing is shown in Figure 6.

Table 1: Forward excursion reduction due to bracing

Severity	Region	Excursion Reduction (%)	p-value
Low	Knees	50	**0.009**
	Pelvis	60	**0.012 – 0.013**
	Elbows	35 – 60	**0.001 – 0.012**
	Shoulders	70	**0.001 – 0.006**
	Head	47	**0.001**
Medium	Knees	18 – 26	0.069 – 0.210
	Pelvis	25	0.083 – 0.109
	Elbows	64	**0.001**
	Shoulders	68	**0.001 – 0.003**
	Head	36	**0.007**

Note: Boldface font indicates statistical significance.

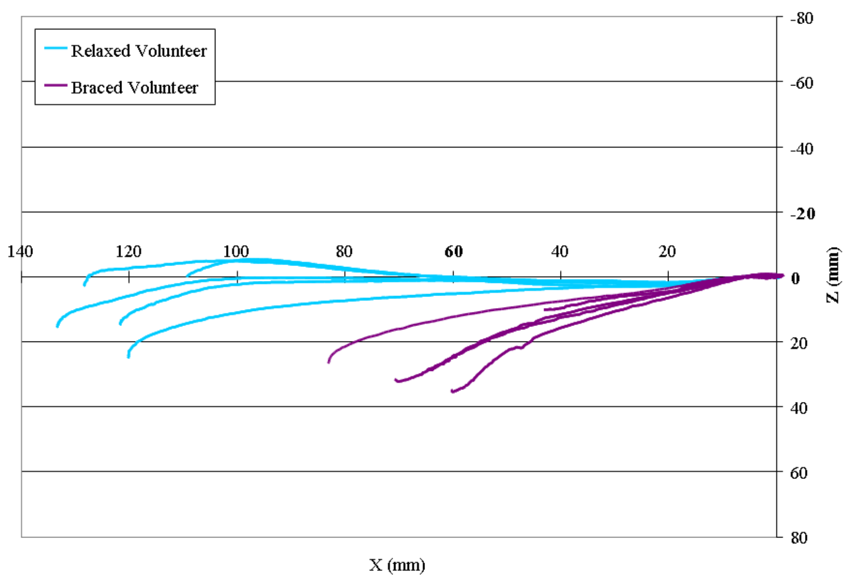

Figure 6: Representative comparison plot of relaxed and braced volunteer excursions. Note: Normalized head CG excursion.

The forward excursions of the ATD were compared to the forward excursions of both relaxed and braced volunteers (Table 2, Table 3). Peak forward excursion of the ATD was, in general, more representative of a braced volunteer than a relaxed volunteer.

Table 2: Relaxed volunteer - ATD comparison

Comparison	Severity	Region	p-value
ATD vs. Relaxed Volunteer	Low	Knees	**0.000**
		Pelvis	**0.002**
		Elbows	**0.000 – 0.010**
		Shoulders	**0.001 – 0.007**
		Head	**0.001**
	Medium	Knees	**0.000 – 0.002**
		Pelvis	**0.005 – 0.026**
		Elbows	0.032 – 0.073
		Shoulders	**0.006 – 0.027**
		Head	**0.013**

Note: Boldface font indicates statistical significance.

Table 3 Braced volunteer - ATD comparison

Comparison	Severity	Region	p-value
ATD vs. Braced Volunteer	Low	Knees	0.221 – 0.287
		Pelvis	0.748 – 0.834
		Elbows	0.034 – 0.693
		Shoulders	**0.002 – 0.021**
		Head	**0.010**
	Medium	Knees	0.138 – 0.258
		Pelvis	0.461 – 0.527
		Elbows	**0.000 – 0.005**
		Shoulders	**0.003 – 0.007**
		Head	**0.013**

Note: Boldface font indicates statistical significance.

CONCLUSIONS

This study illustrates that muscle activation has a significant impact on the biomechanical response of human occupants in low-speed frontal impacts. The kinematic analysis revealed that forward excursions of select anatomical regions were considerably reduced when muscles were activated prior to impulse initiation. It is also of note that the percent decrease in forward excursion due to muscle tension decreased with increasing impulse severity. These findings further reflect the importance of incorporating dynamic muscle activation in computational models and ATDs. With regard to the ATD tests, the peak forward excursion of the current Hybrid III 50th percentile male ATD is more representative of a braced volunteer than a relaxed volunteer. Overall, this study provides novel biomechanical data that can be used to refine and validate computational models and ATDs used to assess injury risk in automotive collisions.

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