Analysis of Force-Limiting Capabilities of Football Neck Collars

David Eugene McNeely III

Thesis submitted to the faculty of the
Virginia Polytechnic Institute and State University
in partial fulfillment of the requirements for the degree of

Master of Science

in

Biomedical Engineering

Stefan M. Duma, Ph.D., Chair
P. Gunnar Brolinson, D.O.
H. Clay Gabler, Ph.D.

May 1, 2006

Blacksburg, Virginia

Keywords: stinger, burner, brachial plexus, football, neck injury

Analysis of Force-Limiting Capabilities of Football Neck Collars

David Eugene McNeely III

ABSTRACT

The purpose of this study was to examine football neck collars and determine their effectiveness at preventing transient brachial plexopathy and other neck injuries due to football impacts. Transient brachial plexopathy, commonly called a stinger or burner, is an injury to the brachial plexus. As many as 65% of collegiate football players will receive suffer such an injury. Accessory neck collars are worn to mitigate the risk of stingers, although little research has been performed to test their effectiveness. In addition to the standard shoulder pad and helmet combination, three collars were tested: the McDavid Cowboy Collar, a collar designed by a Virginia Tech physician called the Bullock Collar, and a prototype device called the Kerr Collar. This study utilized a Hybrid-III 50th percentile male outfitted with a standard collegiate football helmet and shoulder pads, and impacted with a linear pneumatic impactor. Forty eight total impacts were performed; impacts were performed at side, front, and axial loading impact locations, with low and high speed impacts, and normal and raised shoulder pad configurations. Each collar was effective at some positions, but no collar was effective at all impact locations. The Cowboy Collar reduced lower neck bending moments in the front position, but raised upper neck bending moments. It also reduced lower neck bending moments in the side position, but only in the raised configuration. The Bullock

Collar was effective at reducing lower neck bending moment in the side position. The Kerr Collar was effective at reducing lower neck bending moments in the side impact location, and provided a larger percent reduction in impactor force in the axial loading position, compared to the shoulder pads alone. Further testing is needed at lower impact velocities that more closely represent injurious impacts in the field.

ACKNOWLEDGEMENTS

First, I would like to thank my parents for always giving me every opportunity to succeed – I couldn't ask for anything more, and I owe my success to them. Thanks as well to my brother Wil for always keeping me grounded and offering support and healthy competition.

Thanks to my advisor Dr. Stefan Duma, who has provided the guidance, resources, and contacts that made this work possible, as well as everyone at the Center for Injury Biomechanics. Dr. Gunnar Brolinson provided much advice and counseling during the course of my tenure here. Also, thank you to Dr. Clay Gabler for serving on my committee and offering a unique perspective.

I would also like to think David Halstead, Scott Halstead, and Mel Cook at Southern Impact Research Center, who introduced me to injury biomechanics.

Table of Contents

List of Figures

List of Tables

INTRODUCTION

Upper trunk brachial plexopathy, commonly called a stinger or burner, is extremely prevalent in competitive football. Studies have found lifetime injury incidences from 49 to 65% in college football teams (Clancy 1977, Sallis 1992). This injury, while usually transient in nature, has the potential to develop into a more serious condition over time (Clancy 1977, Speer 1990). Accessory neck collars have been implemented in the past to prevent this injury, but research on these collars is limited to a few quasi-static tests, and no dynamic impact testing.

Injury Incidence and Mechanisms

A stinger injury most likely affects the upper trunk of the brachial plexus, which is made up of the C5 and C6 nerve roots (Robertson 1979). This group of nerves runs from the cervical spine through the shoulder and into the upper arm, traveling directly under the clavicle (Figure 1). Stingers usually involve excessive hyperextension or lateral flexion of the head due to an impact, either with another player or with the ground. Symptoms include numbness, pain, or a stinging or burning sensation in the shoulder and arm. Symptoms usually resolve within minutes (Clancy 1977). However, this simple neuropraxia can escalate into an axonotmesis (damage to the axon or myelin sheath) that lasts for days or months, or a neurotmesis (complete disruption of the nerve) that is permanent (Hershman 1990).

There are two main lateral flexion injury mechanisms: traction and compression. In a traction injury, the head is flexed laterally, and the brachial plexus ipsilateral to the impact is stretched. In a compression injury, lateral flexion of the head leads to a pinching of the nerve roots when the foramina close on the contralateral side (Sallis 1992) (Figure 2, Figure 3). This type of injury is usually very precise and local, while the stretching injury may occur anywhere along the plexus and is usually a more diffuse injury.

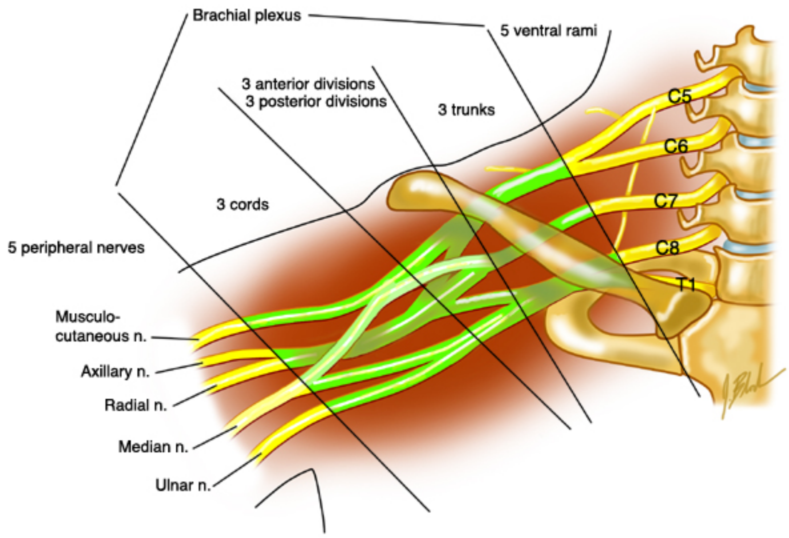

Figure 1: Brachial plexus (photo: http://www.backpain-guide.com/Chapter_Fig_folders/Ch05_Anatomy_Folder/Ch5_Images/05-9_Brachial_Plexus.jpg. Used with permission.)

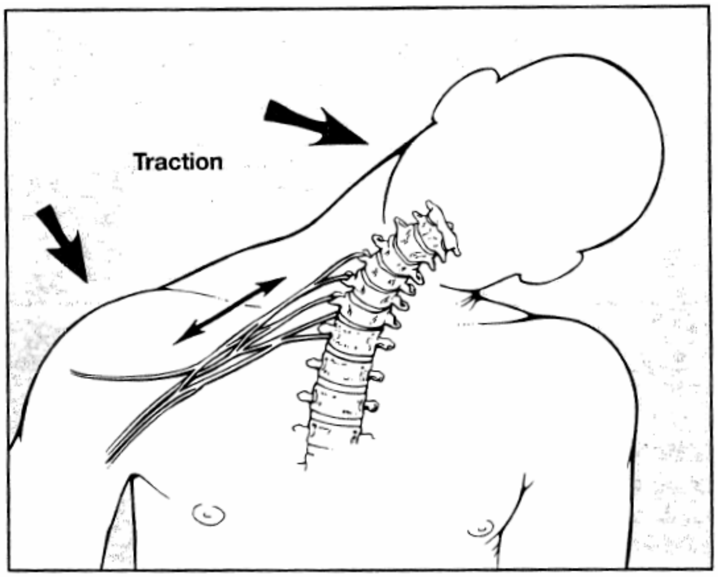

Figure 2: Stinger injury mechanism, traction (Sallis 1992. Used with permission.)

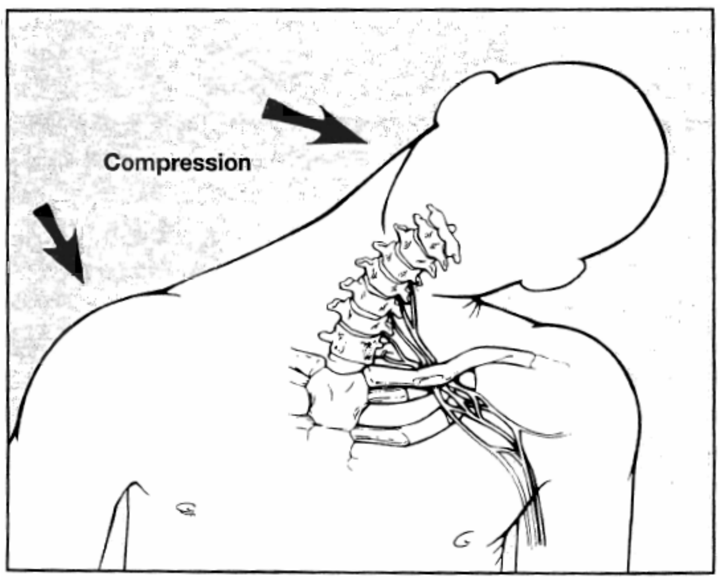

Figure 3: Stinger injury mechanism, compression (Sallis 1992. Used with permission.)

Severe injuries often result from axial loading injuries. When the neck is flexed 30 degrees from anatomic position, the normal cervical lordosis is straightened and the vertebrae align into a segmented column. An impact to the head will result in a crushing of the fragile vertebrae, with the surrounding soft tissues unable to absorb the impact

(Torg 1990). Such impacts usually result in paralysis or death. During 1971 and 1975, the National Football Head and Neck Injury Registry recorded 259 cervical spine fractures, subluxations, and dislocations in high school and college football. Because of the high incidence of such serious injuries, American football instituted rule changes outlawing head-first tackling, blocking, and spearing in 1976. Since that time, the incidence of severe cervical injuries has plummeted; in 1987, 32 injuries were recorded (Torg 1990). This injury is still one of concern, but the aforementioned rule changes combined with coach and player education have greatly reduced such injuries.

Characterization of neck injury is difficult; soft tissue injuries are often hard to diagnose and pose a challenge for researchers to replicate. One way of characterizing the severity of an impact as it relates to injury of the neck is the Neck Injury Criterion, or Nij, which was developed by NHTSA as a way to evaluate neck injury in motor vehicle accidents (Eppinger 2000). This injury criterion is used to predict tension-extension injuries, tension-flexion injuries, compression-extension injuries, and compression-flexion injuries. The injuries that are likely to produce a stinger are tension-extension injuries (Eppinger 2000).

To calculate Nij, data from a Hybrid-III's upper neck load cell is collected. The axial load (Fz) and the flexion/extension bending moment at the occipital condyles (My) are compared to critical intercept values. The formula for calculating Nij is

$$Nij = \frac{Fz}{F_{\text{int}}} + \frac{My}{M_{\text{int}}} \qquad \text{Equation 1}$$

Where Fz is the axial load, F_{int} is the critical intercept value of load used for normalization, My is the flexion/extension bending moment, and M_{int} is the critical intercept value of moment used for normalization. At each instance in time, the Nij is calculated.

Traditional Nij values are useful only for frontal collisions in motor vehicle accidents, and this corresponds to the front impact location in this study. For the side impact location, a modified Nij was used which was developed by Duma (Duma 2003). Instead of My, the lateral bending moment at the occipital condyles, or Mx, and its corresponding critical intercept value are used.

Previous Research

The collars that are worn by football players to prevent this injury were most often designed and put into use without the benefit of scientific scrutiny, and rely heavily on empirical data. Two researchers have attempted to quantify the effectiveness of these collars: Hovis in 1994 and Gorden in 2003.

Hovis and his collaborators outfitted a subject with a helmet and various shoulder pad/collar combinations. The Cowboy Collar, a foam neck roll, and a custom cervical orthosis were tested. A pulley system was used to apply a quasi-static load to the subject's head to produce either hyperextension or lateral flexion of the neck. A consistent bending force was applied by the subject himself. The maximum cervical motion was determined through goniometric analysis, or examination of the calibrated image files. The cervical motion was expressed as a percentage of reduction in hyperextension or lateral flexion as compared to the helmet alone. The shoulder pads provided a percent reduction of 3.52% compared to the helmet alone, while the collars provided reductions of 33.36 to 48.36%, in hyperextension of the neck. The study found no difference in reduction of motion for lateral flexion of the neck (Hovis 1994).

Gorden took a similar approach in analyzing football neck collars, but opted to apply a force with a hand-held pressure transducer. The test subjects were fitted with a helmet and shoulder pads, and a variety of neck collars. The Cowboy Collar, a foam neck roll, and the A-Force Neck Collar were tested. A force of 133.5 N was applied in the anterior-posterior and lateral directions, and the maximum distance traveled was noted using video. In addition, active motion trials were performed, in which the subject moved his head to his maximum hyperextension or lateral flexion. In the front-loading position, the researchers found that all collars permitted significantly less hyperextension than the shoulder pads alone, while the shoulder pads did not significantly reduce motion compared to the helmet alone. In the lateral loading tests, it was found that the collars did not significantly affect the active motion of the head; however, the neck roll permitted significantly less passive motion than the other shoulder pad/brace configurations (Gorden 2003).

Objectives

To date, there has been no dynamic analysis of the effectiveness of football neck collars in preventing hyperextension or lateral flexion of the neck. The overall objectives of this study are to:

- Understand neck loading in football impacts
- characterize the kinematics of injurious impacts to the head and neck
- characterize the kinetics of injurious impacts to the head and neck
- compare existing neck braces
- investigate development of a standard to which neck braces might be tested.

METHODS

A 50% percentile male Hybrid-III anthropomorphic test dummy was utilized. The dummy was fitted with a standard set of Douglas model CP25 shoulder pads, and a Riddell VSR-4 helmet. Impacts were performed with this standard configuration, as well as with one of three accessory collars: the McDavid Cowboy Collar™, a custom-designed and fitted collar worn by a Virginia Tech player called a Bullock Collar, and a prototype device called the Kerr Collar.

Collars

The Cowboy Collar is manufactured by McDavid. It features a molded polyethylene foam collar, and is designed to be laced into the shoulder pads. There is no other anchoring method other than the laces (Figure 4). The Kerr Collar is designed to be worn under the shoulder pads, and features ridges designed to contact the lower edge of the football helmet in an impact. There are rigid stabilizers mounted inside the ridges (Figure 5). The Bullock collar was designed by Dr. Richard Bullock, a head team physician for the Virginia Tech football team for many years. The Bullock Collar is custom-designed and fitted for each specific player. It features a high-density foam collar with a rigid plastic insert. It is attached with straps that are bolted to the shoulder pads (Figure 6).

Figure 4: Cowboy Collar

Figure 5: Kerr Collar

Figure 6: Bullock Collar

The collars, as well as the standard shoulder pads, were evaluated both in a normal state, and in a raised state, which was meant to simulate a player assuming a tackling posture, in which the shoulders are naturally raised in anticipation of an impact. For the standard shoulder pads and the Bullock Collar, foam blocks were inserted underneath the shoulder pads to raise them up until the bottom of the helmet was touching the pads. For the Cowboy Collar and the Kerr Collar, expandable polyurethane foam was poured into bags located under the collars in order to raise them until the collars were touching the bottom of the helmet.

Instrumentation

The impacts were performed with a pneumatic linear impactor (Figure 7). When activated, a solenoid opens a butterfly valve, which sends a blast of compressed air into a chamber. Inside this chamber is a piston, which is accelerated out of the chamber and pushes an impactor arm with a weight of 15 kg. The end of the impactor has a hemispherical nylon shell with high-density vinyl nitrile foam underneath. This impacting surface was designed to replicate the impacting characteristics of a typical football helmet, and is identical to the impacting surface used in the new proposed NOCSAE standard for football helmet testing (Pellman 2006). This impact surface is elastic, and with a minimal rest period between impacts (two minutes), its impact properties do not change over the course of testing. The impactor has a maximum velocity of 15 m/s. Velocities of 7.5 m/s and 11 m/s were used for this study. Impact velocities were controlled to within 3.4% error. These impact velocities were chosen based upon a review of the new proposed NOCSAE standard for helmet certification (Pellman 2006).

Figure 7: Linear pneumatic impactor

The dummy was fitted with three single-axis orthogonally mounted accelerometers in the center of gravity of the head, and three orthogonal angular rate sensors. Upper and lower six-axis load cells were also used. Chin strap load cells were attached to the chin strap of the helmet. The impactor was fitted with a three-axis load cell and accelerometer, as well as a light gate to record impact velocity. High-speed video was recorded of each impact using a Phantom color video camera operating at 1,000 frames per second (Figure 8). All data was filtered to CFC 180.

The helmet was fitted onto the dummy using a helmet positioning tool that uses landmarks on the helmet and the dummy's face to position the helmet. The tool is aligned with the dummy's nose and the helmet is adjusted until the edge of the facemask is aligned with a line on the tool (Figure 9). The dummy was positioned relative to the impactor by aligning a target on the helmet with the center of the impactor (Figure 10).

With these tools, the positioning of the helmet on the dummy is consistent, as well as the positioning of the dummy relative to the impactor. Each of these can be controlled to within 3 degrees.

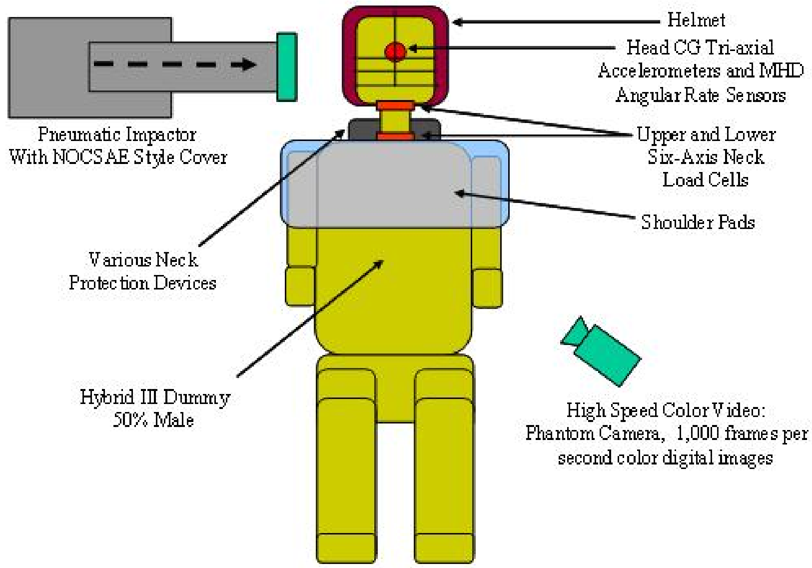

Figure 8: Instrumentation setup

Figure 9: Helmet positioning tool

Figure 10: Center of impactor marker

Impact Locations

Three impact locations were used: a side location, which was located approximately on the Frankfurt plane directly on the side of the head a front location, which was located directly above the uppermost edge of the facemask and an axial loading location, which was chosen to create a situation in which the neck acts as a segmented column with a minimum of bending (Appendix B). The axial loading condition was located approximately halfway from the front position and a crown impact on the top of the head. The impact locations were marked on the helmet, and a helmet positioning tool was used to ensure that the helmet was positioned on the dummy head in the same fashion for each impact (Appendix B). The test matrix yielded a total of 48 impacts.

RESULTS

Data were collected for 48 impacts. For each impact, head acceleration and angular velocity, upper and lower neck loads and moments, impactor loads, and impact velocity was recorded, as well as high-speed video (Figure 11). Data can be found in Tables 1-4 and Appendix A. Figures A 19, A 20, and A 21 illustrate that impactor load and bending moments are independent of impact velocity.

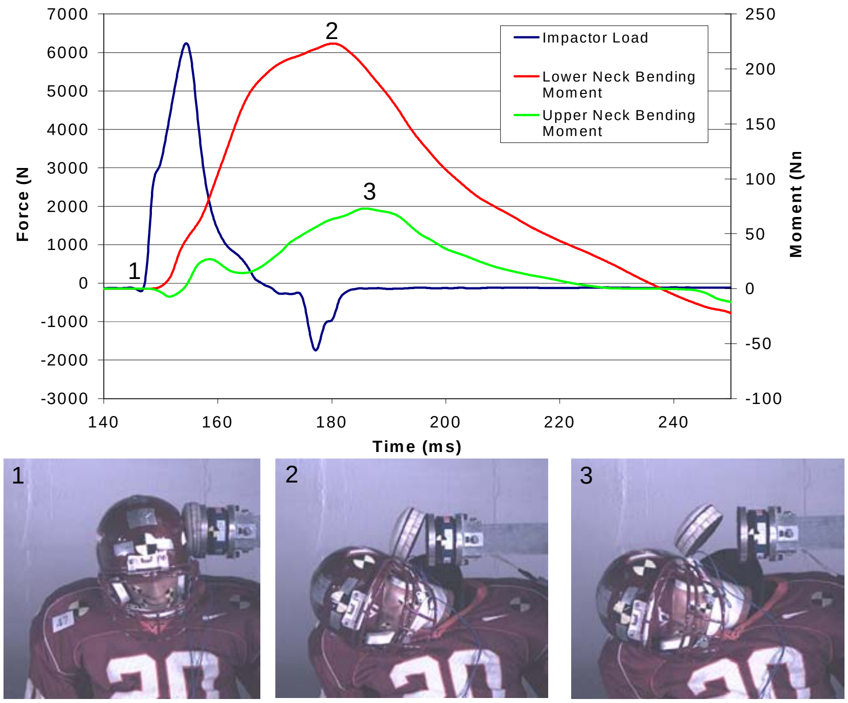

Figure 11: Time traces for exemplar impact. Images correspond to first contact (1), peak lower neck bending moment (2), and peak upper neck bending moment (3).

<u>Side Impact Location</u>

Results for the side impact location are presented. We are most concerned with the lower neck bending moment in the lateral direction, Mx. For the control situation with only the shoulder pads and helmet, the lower neck load cell recorded a peak moment of 145.06 N·m at the low speed impact, and 222.54 N·m for the high speed impact. The moments with the Cowboy Collar were virtually the same, with 145.84 N·m at the low speed and 223.11 N·m at the high speed. The Kerr Collar reduced the lateral bending moment slightly, providing 135.06 N·m at the low speed and 188.29 N·m at the high speed impacts. Finally, the Bullock Collar also reduced this moment, with 142.98 N·m and 200.60 N·m at the low and high speed impacts respectively (Figure 12).

All shoulder pad/collar combinations performed better in the raised positions; the control configuration provided 139.58 N·m and 188.82 N·m in the low and high speed impacts. The Cowboy Collar, in the raised position, was able to reduce these moments to 121.45 N·m and 167.25 N·m for the low and high speed impacts. The Kerr and Bullock Collars had a smaller effect, with 126.06 N·m and 178.34 N·m for the low and high speeds for the Kerr, and 125.83 N·m and 184.76 N·m for the Bullock Collar (Figure 13).

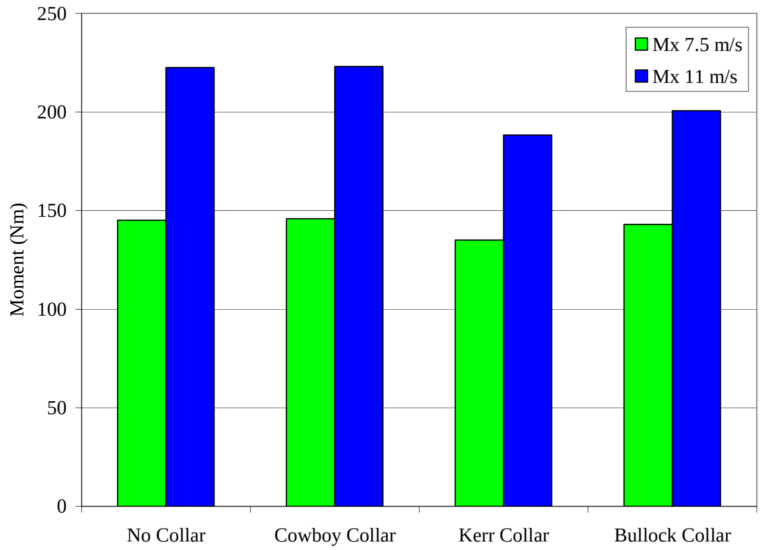

Figure 12: Side impact location, lower beck bending moment

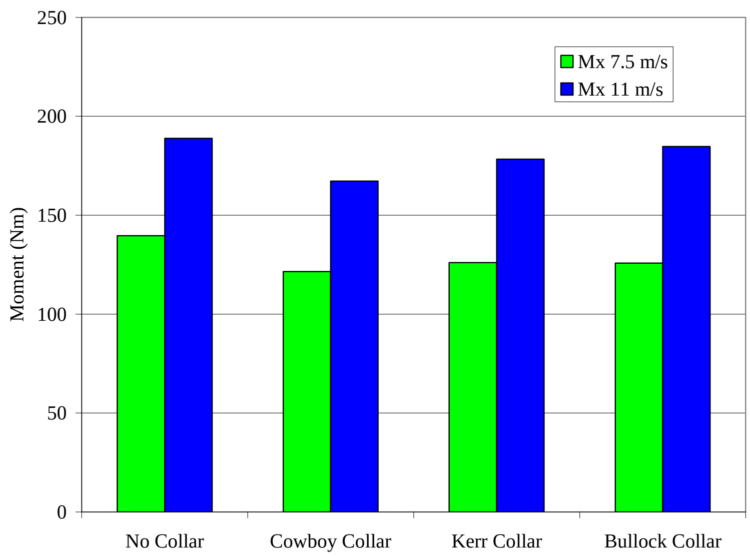

Figure 13: Side impact location, raised shoulder pads, lower neck bending moment

Front Impact Location

In the front location, we examine the lower neck bending moment in the anterior-posterior direction, My. The control configuration provided lower neck bending moments of 123.28 N·m for the low speed and 284.10 N·m for the high speed impact. The Cowboy Collar provided a slight reduction in these forces for the low speed impact with 127.14 N·m, but was the best performer in the high speed impact with 226.11 N·. The Kerr Collar provided a reduction in both tests with 121.72 N·m and 254.98 N·m in the low and high speed impacts, respectively. The Bullock Collar proved to be the best performer in the low speed impact at 116.99 N·m, and reduced the moment to 254.72 N·m in the high speed impact (Figure 14).

It is interesting to note that these moment reductions at the lower neck generally correlate to higher moments in the anterior-posterior bending moment of the upper neck. The control configuration provided 63.75 N·m for the low speed impact and 81.52 N·m for the high speed impact. The Cowboy Collar raised these numbers to 67.06 N·m for the low speed impact, and provided the largest increase to 140.67 N·m for the high-speed impact. The Kerr Collar's tests resulted in 74.19 N·m and 104.21 N·m for the low and high speed impacts. The Bullock Collar performed similarly, with 73.39 N·m at the low speed impact and 114.47 N·m at the high speed impact (Figure 15).

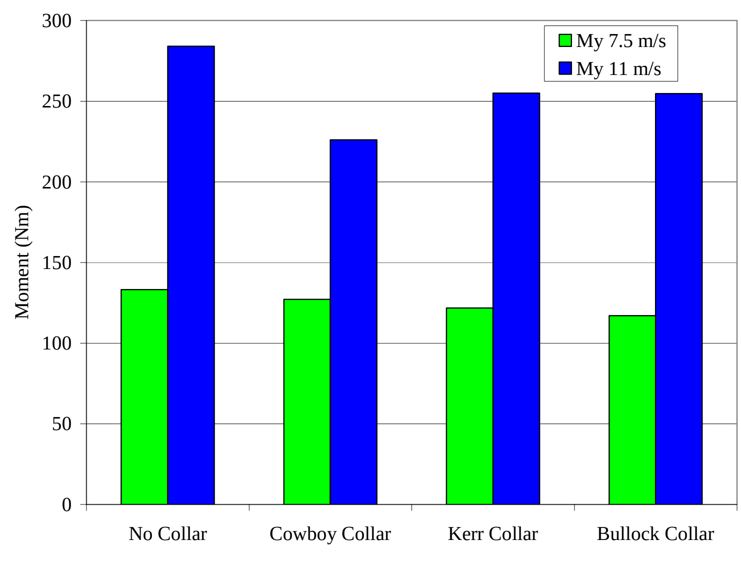

Figure 14: Front impact location, lower neck bending moment

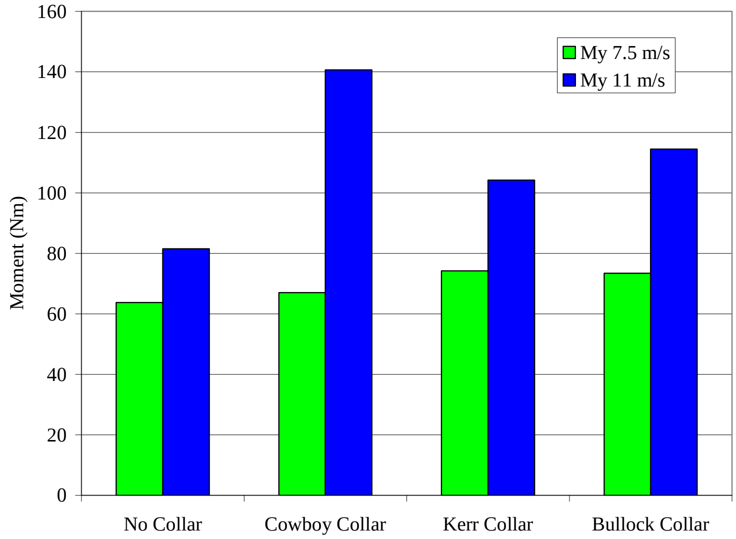

Figure 15: Front impact location, upper neck bending moment

<u>Axial Loading Impact Location</u>

At the axial loading location, we examined the percent reduction of impactor load. The axial load at the lower neck load cell, Fz, is considered as a percentage of the impactor load. The control configuration's tests resulted in a 14.39% reduction of impact load for the low speed impact, and a 24.41% reduction for the high speed impact. The Cowboy Collar provided a larger load reduction, with 18.67% and 27.89% reductions at the low and high speeds respectively. The Kerr Collar proved to be the best performer, with a 23.23% reduction at the low speed impact and a 28.78% reduction at the high speed impact. The Bullock Collar also provided a larger percent reduction, with a 16.01% reduction at the low speed, and a 22.38% reduction at the high speed (Figure 16).

When the collars and shoulder pads were raised, the results differed slightly. The control configuration provided a 16.38% reduction at the low speed, and a 28.18% reduction at the high speed. The Cowboy Collar performed worse than the control, with 11.31% and 23.41% reductions at the low and high speeds. The Kerr Collar provided a larger percent reduction in both tests, with 28.25% and 32.73% reductions at the low and high speeds. The Bullock Collar provided a small reduction at the low speed with 17.97%, but performed worse than the control at the high speed, with 26.46% reduction (Figure 17).

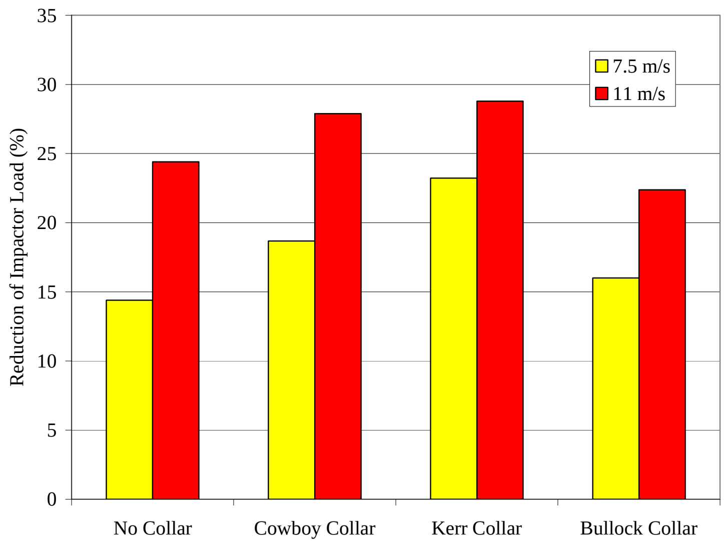

Figure 16: Axial loading impact location, percent reduction of impactor load

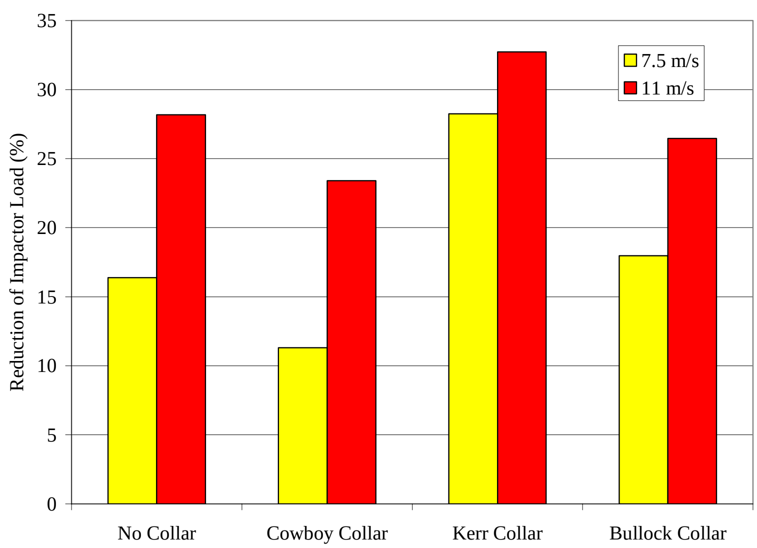

Figure 17: Axial loading impact location, raised shoulder pads, percent reduction of impactor load

Table 1: Data from impact tests (page 1 of 4)

Test Number	Impact Location	Collar Type	Shoulder Position	Target Velocity (m/s)	Actual Velocity (m/s)	Percent Error
neck34	side	No Collar	normal	7.5	7.5	0.1
neck35	side	No Collar	normal	11	11.0	0.4
neck36	front	No Collar	normal	7.5	7.6	1.6
neck37	front	No Collar	normal	11	11.1	1.1
neck38	AL	No Collar	normal	7.5	7.4	1.4
neck39	AL	No Collar	normal	11	11.1	1.1
neck40	AL	No Collar	raised	7.5	7.5	0.1
neck41	AL	No Collar	raised	11	10.7	2.4
neck42	front	No Collar	raised	7.5	7.7	2.6
neck43	front	No Collar	raised	11.0	10.8	1.8
neck44	side	No Collar	raised	7.5	7.6	1.6
neck45	side	No Collar	raised	11.0	11.4	3.4
neck46	side	Cowboy Co	normal	7.5	7.7	2.6
neck47	side	Cowboy Co	normal	11.0	11.3	2.6
neck48	front	Cowboy Co	normal	7.5	7.6	1.6
neck49	front	Cowboy Co	normal	11.0	11.2	1.9
neck50	AL	Cowboy Co	normal	7.5	7.6	1.6
neck51	AL	Cowboy Co	normal	11.0	11.0	0.4
neck52	AL	Cowboy Co	raised	7.5	7.4	0.9
neck53	AL	Cowboy Co	raised	11.0	11.0	0.4
neck54	front	Cowboy Co	raised	7.5	7.7	3.1
neck55	front	Cowboy Co	raised	11.0	11.2	1.9
neck56	side	Cowboy Co	raised	7.5	7.6	1.1
neck57	side	Cowboy Co	raised	11.0	11.2	1.9
neck58	side	Kerr Collar	normal	7.5	7.6	1.1
neck59	side	Kerr Collar	normal	11.0	11.0	0.3
neck60	front	Kerr Collar	normal	7.5	7.6	1.6
neck61	front	Kerr Collar	normal	11.0	11.1	1.1
neck62	AL	Kerr Collar	normal	7.5	7.5	0.6
neck63	AL	Kerr Collar	normal	11.0	10.9	1.1
neck64	AL	Kerr Collar	raised	7.5	7.5	0.4
neck65	AL	Kerr Collar	raised	11.0	11.0	0.3
neck66	front	Kerr Collar	raised	7.5	7.7	2.1
neck67	front	Kerr Collar	raised	11.0	10.9	1.1
neck68	side	Kerr Collar	raised	7.5	7.5	0.6
neck69	side	Kerr Collar	raised	11.0	11.0	0.4
neck70	side	Bullock Col	normal	7.5	7.6	1.6
neck71	side	Bullock Col	normal	11.0	11.1	1.1
neck72	front	Bullock Col	normal	7.5	7.8	3.7
neck73	front	Bullock Col	normal	11.0	11.3	2.6
neck74	AL	Bullock Col	normal	7.5	7.5	0.1
neck75	AL	Bullock Col	normal	11.0	11.0	0.3
neck76	AL	Bullock Col	raised	7.5	7.3	2.3
neck77	AL	Bullock Col	raised	11.0	11.0	0.3
neck78	front	Bullock Col	raised	7.5	7.5	0.1
neck79	front	Bullock Col	raised	11.0	11.0	0.4
neck80	side	Bullock Col	raised	7.5	7.7	2.1
neck81	side	Bullock Col	raised	11.0	11.2	1.9

Table 2: Data from impact tests (page 2 of 4)

	Lower Neck Loads (N)				Lower Neck Moments (Nm)			
Test Number	**Fx**	**Fy**	**Fz**	**Res**	**Mx**	**My**	**Mz**	**Res**
neck34	149	501	532	639	145	57	13	155
neck35	229	1120	882	1316	223	60	33	231
neck36	676	200	3123	3133	25	133	10	134
neck37	995	324	5806	5815	33	284	14	285
neck38	573	230	4496	4503	20	255	12	255
neck39	807	344	8201	8213	48	445	19	446
neck40	591	157	4371	4381	15	239	11	239
neck41	840	229	9562	9566	28	465	16	466
neck42	743	150	2999	3006	11	125	6	125
neck43	1384	130	5087	5089	9	238	7	238
neck44	235	602	615	826	140	33	14	142
neck45	213	1061	1002	1461	189	33	23	190
neck46	124	598	495	754	146	24	18	148
neck47	141	1056	1086	1441	223	55	33	224
neck48	650	104	3021	3025	9	127	6	127
neck49	1393	249	4930	4931	23	226	10	226
neck50	629	157	4410	4419	16	246	14	246
neck51	891	179	8700	8709	18	464	17	464
neck52	518	261	4314	4328	18	231	13	232
neck53	739	245	9249	9252	32	467	12	468
neck54	673	210	2738	2753	15	105	8	106
neck55	961	295	5390	5397	25	254	12	256
neck56	296	525	603	812	121	23	12	122
neck57	335	845	992	1321	167	32	16	168
neck58	254	504	478	703	135	40	16	141
neck59	254	1042	988	1436	188	38	26	192
neck60	695	174	2967	2978	13	122	7	123
neck61	1006	247	5344	5350	24	255	11	256
neck62	606	202	4039	4044	18	220	12	220
neck63	1044	254	8823	8829	27	465	14	466
neck64	487	160	3729	3738	15	200	10	200
neck65	1011	288	8845	8855	24	466	17	467
neck66	649	122	2703	2710	20	111	7	111
neck67	1184	229	4627	4631	28	212	10	213
neck68	205	565	745	941	126	38	16	128
neck69	303	1005	1296	1651	178	36	32	184
neck70	254	632	458	785	143	49	19	152
neck71	260	1092	874	1335	201	59	26	207
neck72	708	133	2873	2883	12	117	14	117
neck73	978	219	5361	5365	14	255	10	255
neck74	414	153	4267	4275	12	238	7	238
neck75	615	229	8315	8322	26	456	13	457
neck76	545	152	4033	4040	11	221	7	221
neck77	666	376	8680	8691	29	464	18	465
neck78	589	151	2948	2957	24	123	10	124
neck79	1056	224	5548	5552	23	265	13	265
neck80	167	562	637	832	126	28	12	126
neck81	270	1063	1092	1520	185	31	28	187

Table 3: Data from impact tests (page 3 of 4)

	Upper Neck Loads (N)				Upper Neck Moments (Nm)			
Test Number	Fx	Fy	Fz	Res	Mx	My	Mz	Res
neck34	118	724	614	769	29	7	8	29
neck35	135	845	1386	1540	64	10	10	65
neck36	704	264	3634	3702	5	64	5	64
neck37	1069	479	6777	6877	16	82	10	82
neck38	309	127	5334	5340	10	53	2	53
neck39	442	147	9862	9870	28	112	3	113
neck40	319	153	5171	5177	6	72	3	72
neck41	516	356	11685	11696	18	130	3	130
neck42	747	87	3492	3558	7	73	2	73
neck43	1232	225	5899	6020	13	132	5	132
neck44	199	547	898	1002	42	8	9	43
neck45	177	659	1614	1687	63	14	14	64
neck46	164	511	751	882	48	10	11	49
neck47	218	560	1619	1706	73	15	9	74
neck48	710	131	3511	3580	7	67	3	67
neck49	1223	240	5699	5824	16	141	5	141
neck50	246	125	5239	5244	9	61	3	61
neck51	356	212	10532	10536	26	100	3	101
neck52	262	166	5078	5083	10	39	2	40
neck53	200	314	11220	11223	18	78	3	78
neck54	661	75	3197	3264	15	60	3	61
neck55	1128	211	6255	6352	19	101	4	103
neck56	254	493	961	1032	38	12	8	38
neck57	267	598	1590	1635	60	13	12	60
neck58	175	624	773	919	47	8	10	48
neck59	203	800	1639	1774	62	13	12	62
neck60	747	120	3452	3527	7	74	2	75
neck61	1142	294	6182	6286	8	104	4	104
neck62	309	113	4719	4728	9	47	2	47
neck63	503	317	10621	10631	9	89	2	89
neck64	306	229	4379	4391	7	34	3	35
neck65	544	398	10470	10486	9	81	3	82
neck66	693	196	3124	3203	5	52	2	52
neck67	1206	298	5309	5446	8	92	3	92
neck68	213	541	977	1093	42	7	7	42
neck69	287	799	1707	1875	79	12	12	79
neck70	185	622	720	918	41	7	13	43
neck71	178	705	1526	1632	59	13	13	60
neck72	742	112	3361	3429	9	73	3	74
neck73	1215	275	6221	6332	9	114	5	115
neck74	313	158	5104	5109	6	52	1	52
neck75	445	220	9928	9938	10	87	2	87
neck76	323	129	4742	4749	8	60	3	60
neck77	409	490	10477	10492	13	89	3	89
neck78	674	246	3421	3490	7	56	3	56
neck79	1153	353	6407	6513	17	111	4	112
neck80	106	499	887	950	51	8	11	52
neck81	214	750	1690	1774	66	16	13	66

Table 4: Data from impact tests (page 4 of 4)

Test Number	Imactor Load (N)	Head Acceleration (g)	Lower Neck Load Fz (N)	Nij	Percent Reduction of Impact Force (%)
neck34	3482	67	532	0.21	
neck35	6256	99	882	0.35	
neck36	4410	60	3123	1.07	
neck37	9674	129	5806	1.91	
neck38	5252	42	4496	1.33	14.4
neck39	10848	69	8201	2.76	24.4
neck40	5227	42	4371	1.37	16.4
neck41	13313	84	9562	3.34	28.2
neck42	4578	66	2999	1.05	
neck43	10621	163	5087	1.89	
neck44	3406	62	615	0.28	
neck45	6393	104	1002	0.50	
neck46	3455	69	495	0.29	
neck47	6241	101	1086	0.48	
neck48	4529	62	3021	1.07	
neck49	10689	163	4930	1.97	
neck50	5422	41	4410	1.34	18.7
neck51	12065	66	8700	2.90	27.9
neck52	4865	39	4314	1.31	11.3
neck53	12075	76	9249	2.94	23.4
neck54	4416	67	2738	0.99	
neck55	10928	150	5390	1.85	
neck56	3379	57	603	0.31	
neck57	6403	101	992	0.52	
neck58	3430	63	478	0.27	
neck59	6606	105	988	0.45	
neck60	4579	66	2967	1.06	
neck61	10962	159	5344	1.85	
neck62	5261	34	4039	1.24	23.2
neck63	12390	71	8823	2.85	28.8
neck64	5197	32	3729	1.13	28.2
neck65	13149	84	8845	2.76	32.7
neck66	4647	59	2703	0.93	
neck67	10534	149	4627	1.57	
neck68	3432	60	745	0.31	
neck69	6702	108	1296	0.49	
neck70	3358	59	458	0.24	
neck71	6359	103	874	0.42	
neck72	4539	65	2873	1.02	
neck73	10969	164	5361	1.86	
neck74	5081	38	4267	1.28	16.0
neck75	10713	65	8315	2.62	22.4
neck76	4917	37	4033	1.24	18.0
neck77	11803	70	8680	2.79	26.5
neck78	4351	59	2948	1.01	
neck79	10921	156	5548	1.92	
neck80	3421	57	637	0.31	
neck81	6333	101	1092	0.50	

DISCUSSION

We predict injury due to flexion in many of the high-speed impacts; the majority of these yielded lower neck bending moments that exceed 190 N·m, which is the injury assessment reference value for flexion injury to the neck from NHTSA (Eppinger 2000). In addition, all of the axial loading tests resulted in higher compression loads than the stated 4000 N limit. The collars had very little effect on head acceleration, but we also predict concussion for many of the high-speed impacts (Figure A 13 - Figure A 18) (Eppinger 2000).

In the front impact location, the Cowboy Collar, and to a lesser extent the Kerr and Bullock collars, reduced the lower neck bending moment. However, there was a corresponding increase of the upper neck bending moment for all collars. Upon inspection of the high-speed video, it is apparent that the collars are changing the point about which the head rotates with respect to the torso. With no collar, the head is free to rotate about the base of the neck. However, with a collar, the pivot is raised up to the point at which the collar contacts the helmet, thus transferring these forces into the upper neck (Figure 18).

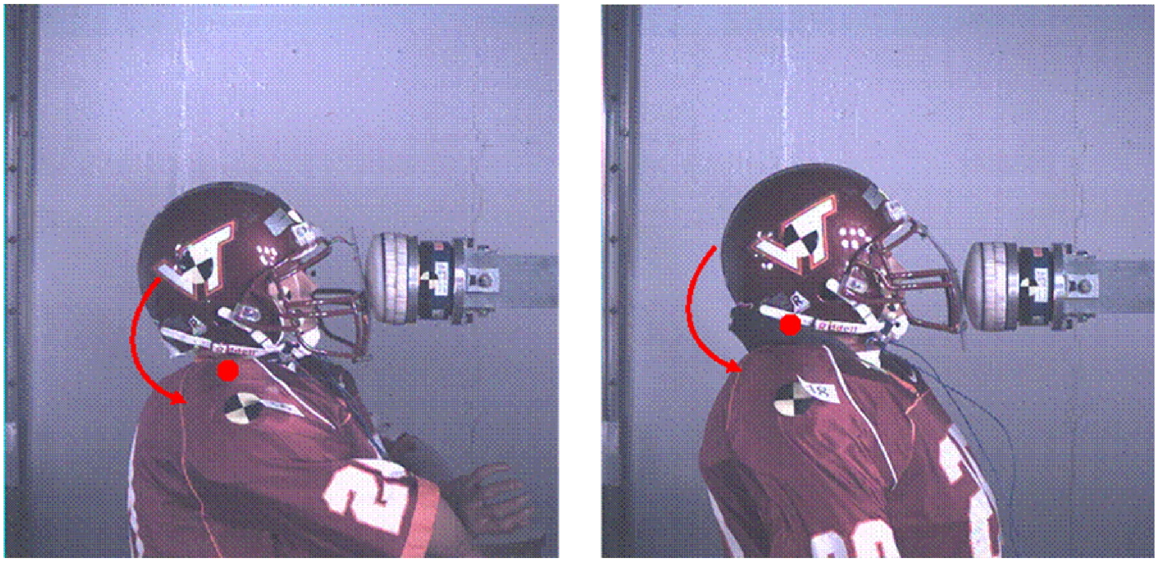

Figure 18: Pivot point with no collar (left), and with Cowboy Collar (right)

Biofidelity of Model

Developing a model of the neck that is biofidelic is a monumental challenge. The anatomy of the cervical spine is quite complicated; the vertebrae and their articulation with each other are intricate (Figure 19). In addition, many small muscle groups provide support and strength of the neck.

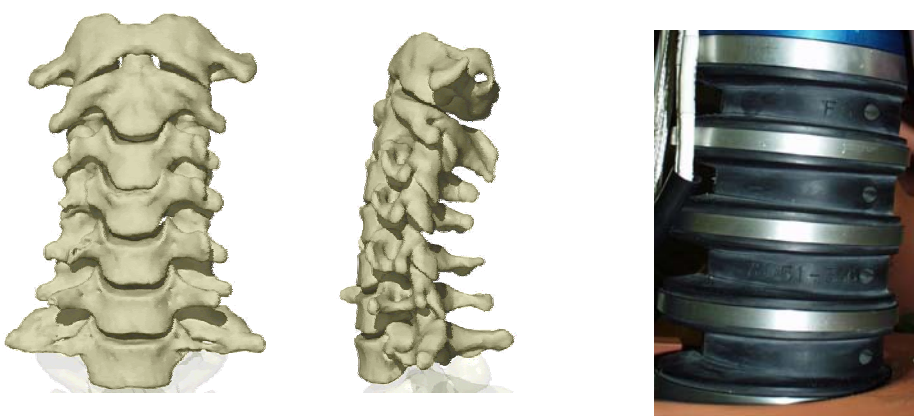

Figure 19: Cervical spine, anterior view (left) and lateral view (center), and Hybrid-III neck assembly (right)

This complex anatomy and musculature presents a unique challenge in creating a mechanical replica that will respond as a human neck would in an impact. The current state of the art in impact testing is the Hybrid-III anthropomorphic test dummy, which was designed for automotive testing. In the Hybrid-III, the neck is simulated with a stainless steel and high-density rubber neck assembly (Figure 19). While the Hybrid-III neck responds similarly to a human's in a frontal crash test, it cannot replicate the complicated kinematics that occur during such impacts, and thus is an imperfect model.

Impact Velocities

Impact velocities for this test were chosen with respect to prior research performed at the Virginia Tech-Wake Forest Center for Injury Biomechanics dealing with concussions, and with consideration of the new proposed NOCSAE standard for helmet testing (Pellman 2006). Upon review of the test results, it has been determined that further testing is needed at lower impact velocities that are similar to injurious impacts in the field with respect to neck injuries. Stingers often occur from glancing or indirect blows that involve low relative impact velocities. Perhaps lower velocities will illustrate differences in the collar's abilities to limit loads and moments in impacts.

Future Testing

A second test series is planned with lower impact velocities. Impact velocities will be chosen to more accurately approximate injurious impacts as they occur in the field. This test series will also incorporate an angular rate sensor array on the Hybrid-III's body, in order to calculate the angle of rotation of the head during an impact.

CONCLUSIONS

Side Impact Location

At the side impact location, it was apparent that none of the collars had sufficient padding to reduce neck loads. Although the Kerr and Bullock collars did reduce the lower neck bending moment slightly, this effect was minimal. When the shoulder pads and collars were raised, the collars were allowed to interact with the helmet to a larger degree and thus were more effective at limiting loads; however, the shoulder pads alone also performed this function, albeit to a lesser extent. It is apparent that there is a lack of padding at the side position of many collars on the market today. Producing a collar that provides support and protection without hindering range of motion is a challenge.

Front Impact Location

At the front impact location, the collars reduced the bending moment at the lower neck, with the Cowboy Collar providing the largest reduction. However, this reduction was always coupled with a corresponding rise the bending moment at the upper neck. Upon analysis of the high-speed video, it appears that the collars are providing a point of rotation that is higher than the control configuration; that is, the collar acts like a fulcrum about which the head rotates. This might lead to a higher risk of injury at the upper neck. When the shoulder pads are raised, this phenomenon is eliminated and the collars are effective at limiting the upper and lower neck bending moments. However, protection while still allowing a free range of motion remains important.

Axial Loading Impact Location

The Kerr Collar proved to be the most efficient collar at providing a reduction of impactor load in the axial loading impact location. The Cowboy Collar and the Bullock Collar also provided a larger percent reduction of load compared to the control configuration; however, when the shoulder pads were raised, the Cowboy and Bullock collars fared worse than the control configuration. Upon analysis of the dummy kinematics, it appears that the collars hold the head and neck in place and force the neck into an axial compression situation. The Kerr Collar provides an even larger percent reduction with the shoulder pads raised.

In summary, all of the collars provided a reduction of loads or moments at some impact locations. The Kerr and Bullock collars provided the largest reduction in bending moment at the side impact location in the normal configuration, while all of the collars reduced this moment in the raised position. The Cowboy Collar, at the raised position, was effective at limiting upper and lower neck loads in the front impact location. The Kerr Collar was effective at reducing the impact load in the axial loading condition.

However, some collars allowed larger loads and moments than the shoulder pads alone. The upper neck loads were higher for all collars in the normal position at the front impact location. In addition, the Cowboy and Bullock collars did not reduce the impactor load as much as the control configuration in the raised axial loading impact location. More investigation is needed to determine if the improper use of these collars will increase the risk of injury in some types of impacts.

REFERENCES

Andrish, J.T., Bergfeld, J.A., and Romo, L. (1977) A method for the management of cervical injuries in football. A preliminary report. American Journal of Sports Medicine. 5(2): 89-92.

Carter, D.R., and Frankel, V.H. (1980) Biomechanics of hyperextension injuries to the cervical spine in football. American Journal of Sports Medicine. 8(5): 302-309

Castro, F.P., Ricciardi, J., Brunet, M.E., Busch, M.T., and Whitecloud, T.S. (1997) Stingers, the Torg ration, and the cervical spine. American Journal of Sports Medicine. 25(5): 603-608.

Clancy, W.G., Brand, R.L., and Bergfield, J.A. (1977) Upper trunk brachial plexus in juries in contact sports. American Journal of Sports Medicine. 5: 209-216.

Duma, S.M., Crandall, J.R., Rudd, R.W., and Kent, R.W. (2003) Small female head and neck interaction with a deploying side airbag. Accident Analysis and Prevention. 35: 811-816.

Duma, S.M., Manoogian, S.J., and Bussone, W.R. (2005) Analysis of real-time head accelerations in collegiate football players. Clinical Journal of Sports Medicine. 15(1): 3-8.

Gorden, J.A., Straub, J.S., Swanik, C.B., and Swanik, K.A. (2003) Effects of football collars on cervical hyperextension and lateral flexion. Journal of Athletic Training. 38(3): 209-215.

Hovis, W.D., and Limbird, T.J. (1994) An evaluation of cervical orthoses in limiting hyperextension and lateral flexion in football. Medicine and Science in Sports and Exercise. 26(7): 872-876.

Markey, K.L., di Benedetto, M., and Curl, W.W. (1993) Upper trunk brachial plexopathy: the stinger syndrome. 21(5): 650-655.

Meyers, S.A., Schulte, K.R., Callaghan, J.J., Albright, J.B., Powell, J.W., Crowley, E.T., and El-Khoury, G.Y. (1994) Cervical spinal stenosis and stingers in collegiate football players. American Journal of Sports Medicine. 22(2): 158-166.

Pellman, E.J., Viano, D.C., Tucker, A.M., and Casson, I.R. (2003) Concussion in professional football: location and direction of helmet impacts part 2. Neurosurgery. 53(6): 1328-1341.

Pellman, E.J., Viano, D.C., Tucker, A.M., Casson, I.R., and Waeckerle, J.F. (2003) Concussion in Professional Football: Reconstruction of game impacts and injuries. Neurosurgery. 53(4): 799-814.

Pellman, E.J., Viano, D.C., Withnall, C., Bir, C.A., and Halstead, P.D. (2006) Concussion in professional football: helmet testing to assess impact performance – part 11. Neurosurgery. 58(1): 78-96.

Sallis, R.E., Jones, K., and Knopp, W. (1992) Burners: offensive strategy for an under-reported injury. The Physician and Sports Medicine**.** 20(11): 47-55.

Speer, K.P., and Bassett, F.H. (1990) The prolonged burner syndrome. American Journal of Sports Medicine. 18(6): 591-594

Torg, J.S., Begso, J.J., O’Neill, M.J., and Sennett, B. (1990) The epidemiologic, pathologic, biomechanical, and cinematographic analysis of football-induced cervical spine trauma. American Journal of Sports Medicine**.** 18(1): 50-57.

Torg, J.S., Sennett, B., Pavlov, H., Leventhal, M.R., and Glasgow, S.G. (1993) Spear tackler’s spine: an entity precluding participation in tackle football and collision activities that expose the cervical spine to axial energy inputs. American Journal of Sports Medicine. 21(5): 640-649.

Virgin, H. (1980) Cineradiographic study of football helmets and the cervical spine. American Journal of Sports Medicine. 8(5): 310-317

APPENDIX A: Impact Data

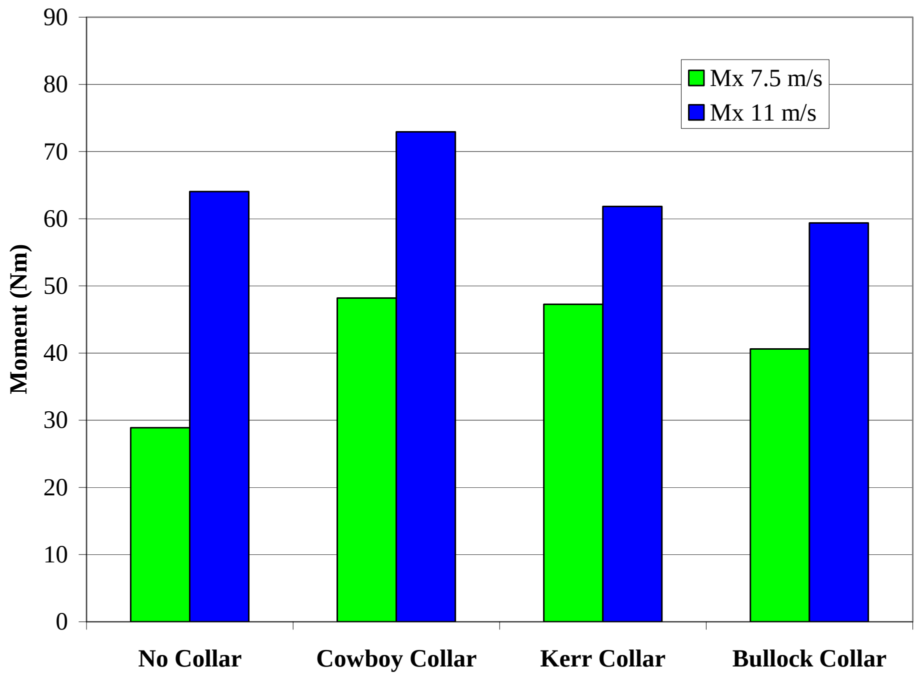

Figure A 1: Side impact location, upper neck bending moment

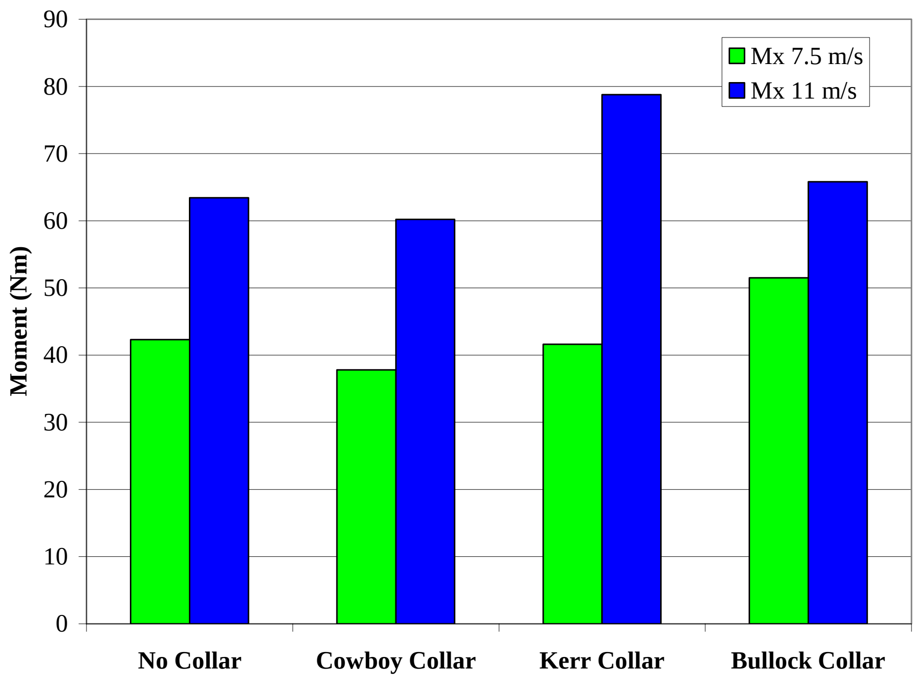

Figure A 2: Side impact location, raised shoulder pads, upper neck bending moment

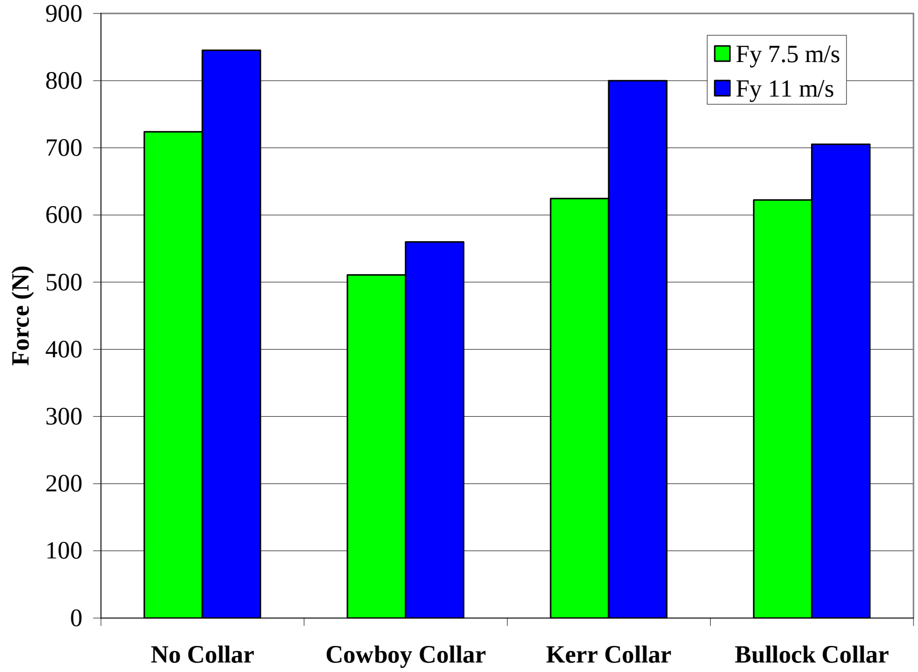

Figure A 3: Side impact location, upper neck shear force

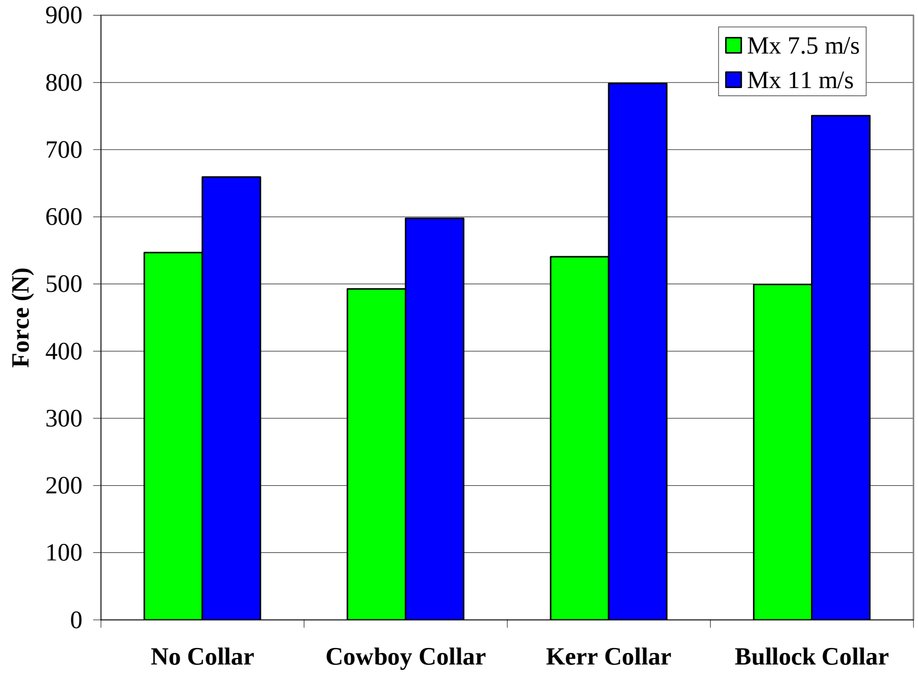

Figure A 4: Side impact location, raised shoulder pads, upper neck shear force

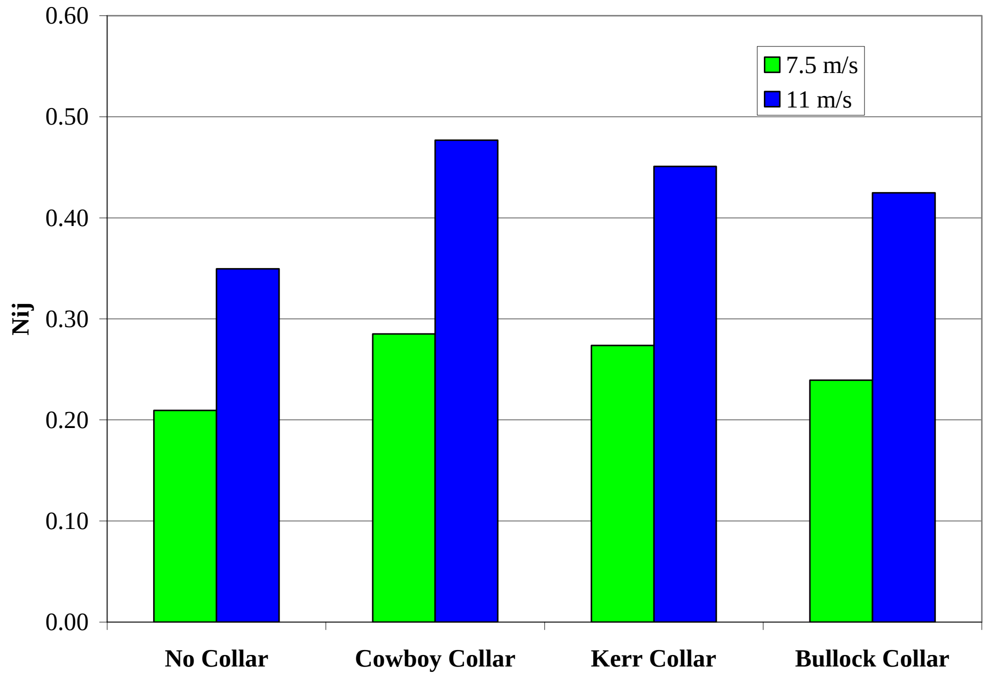

Figure A 5: Side impact location, modified Nij

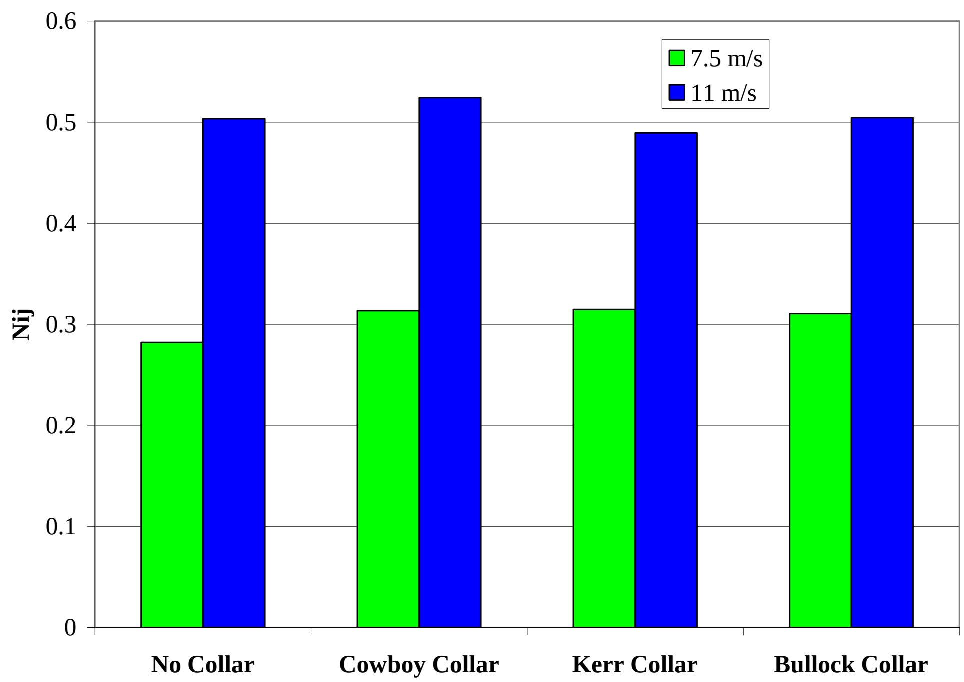

Figure A 6: Side impact location, raised shoulder pads, modified Nij

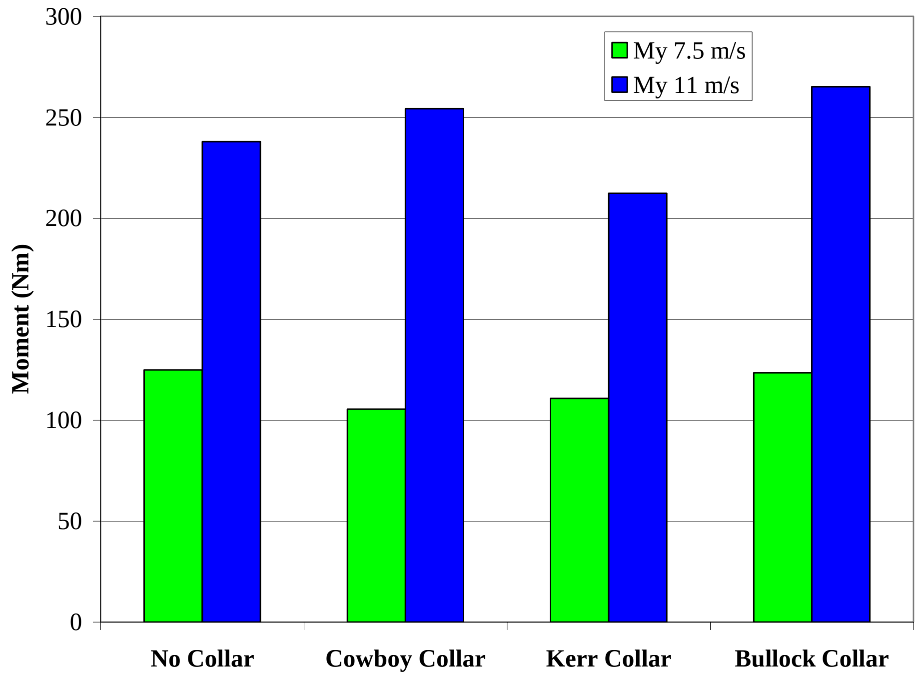

Figure A 7: Front impact location, raised shoulder pads, lower neck bending moment

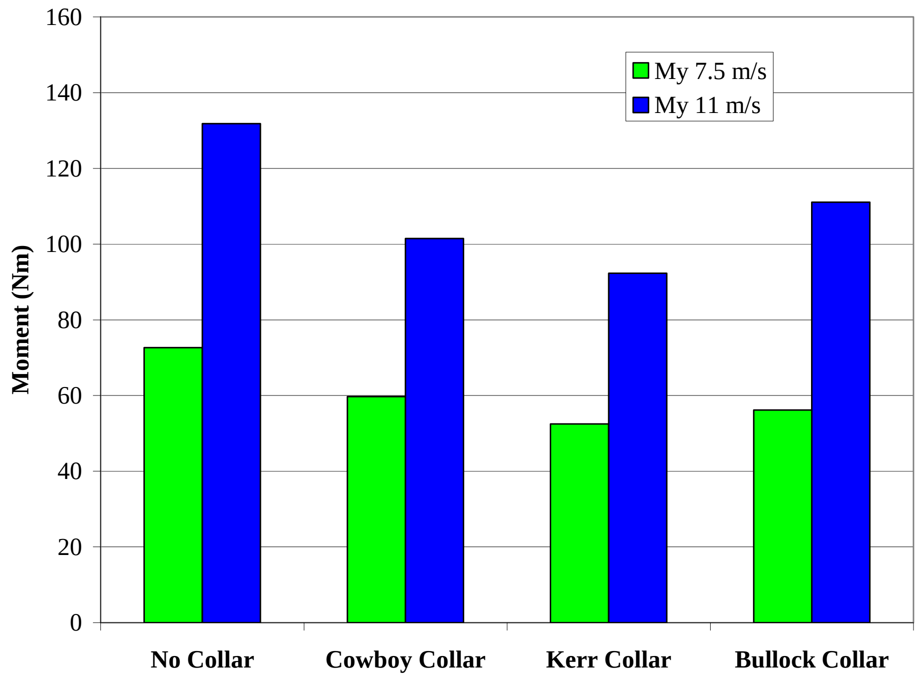

Figure A 8: Front impact location, raised shoulder pads, upper neck bending moment

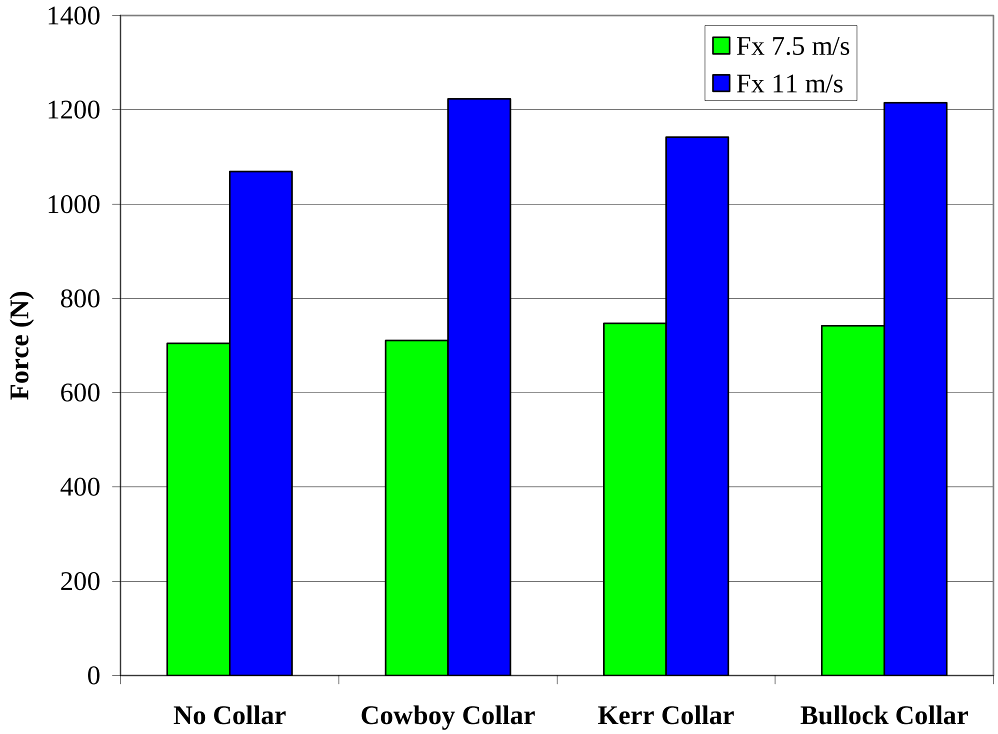

Figure A 9: Front impact location, upper neck shear force

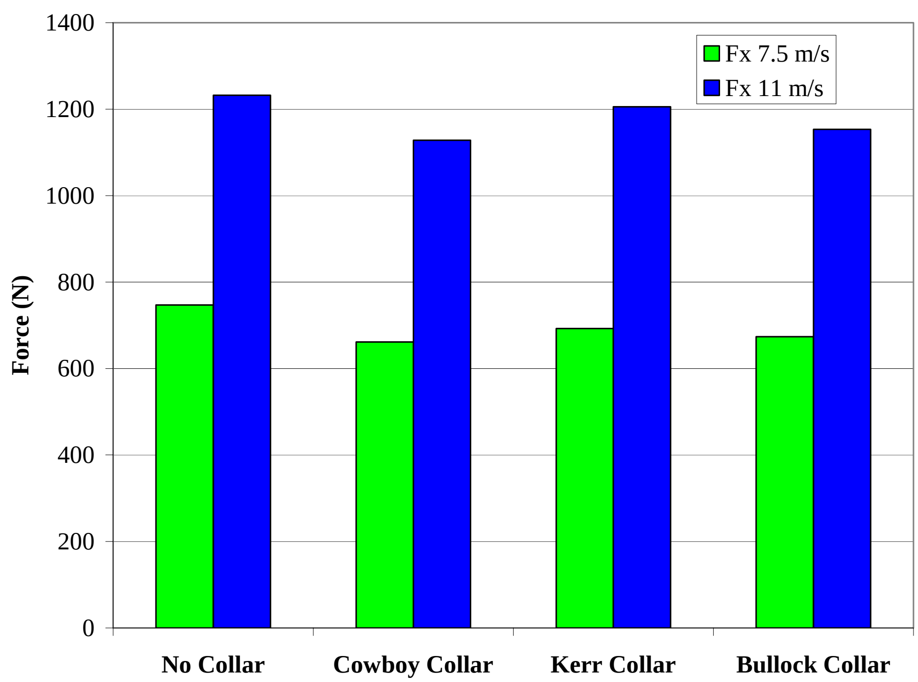

Figure A 10: Front impact location, raised shoulder pads, upper neck shear force

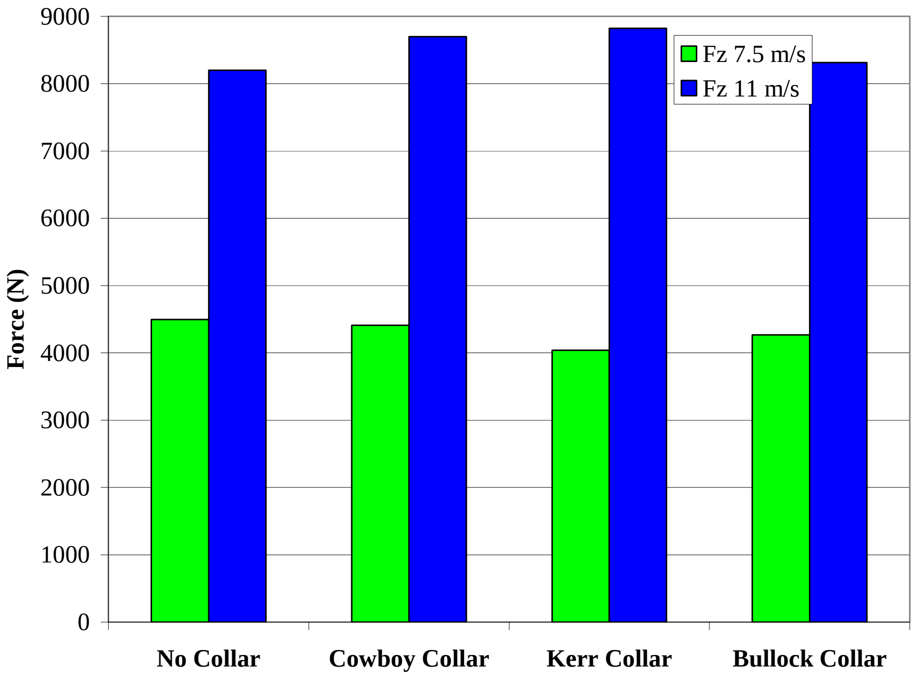

Figure A 11: Axial loading position, lower neck axial compression force

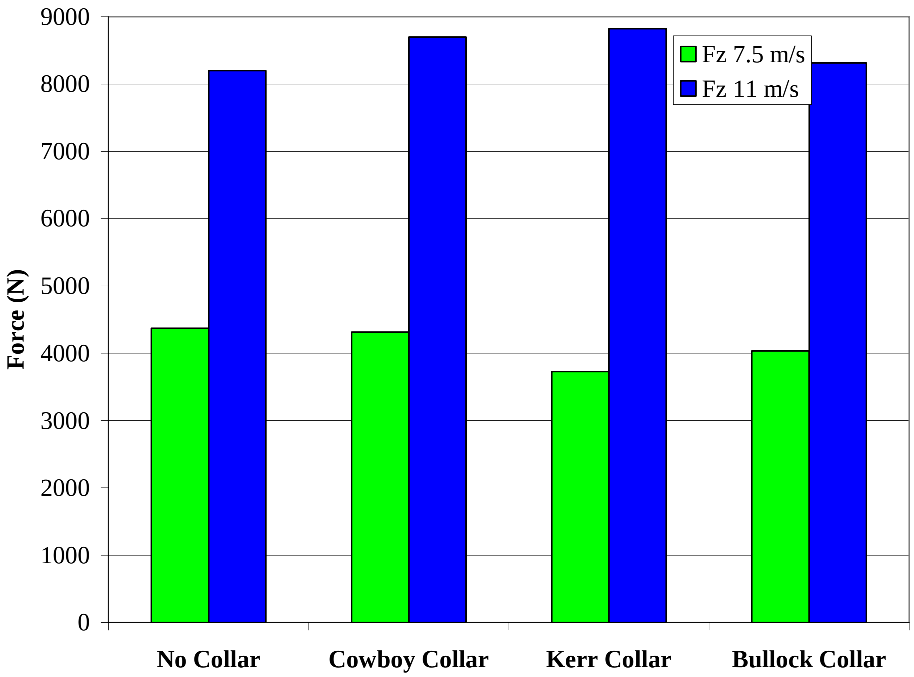

Figure A 12: Axial loading position, raised shoulder pads, lower neck axial compression force

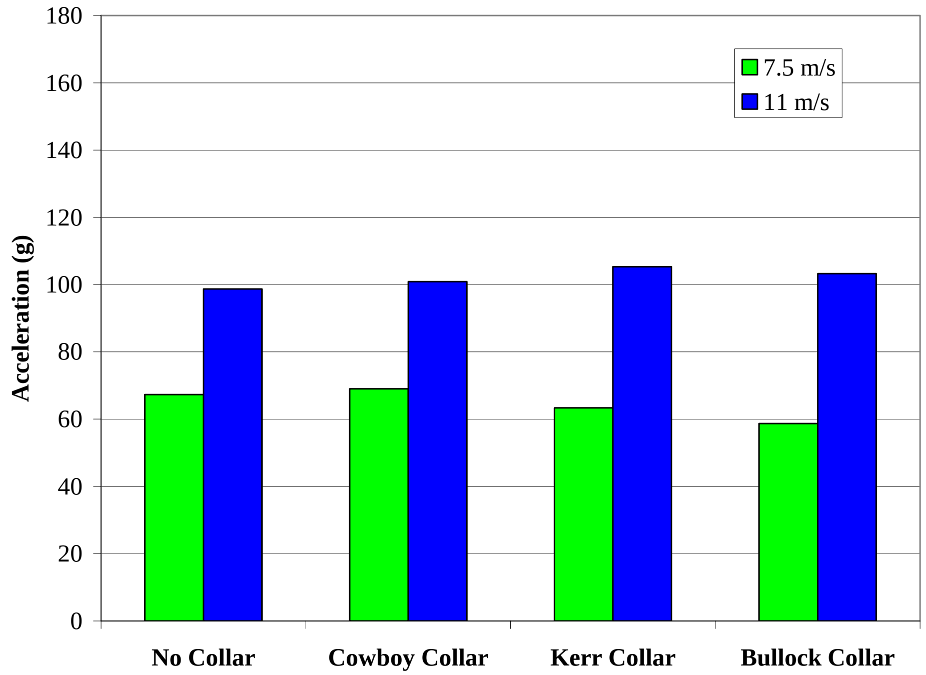

Figure A 13: Side impact location, peak head acceleration

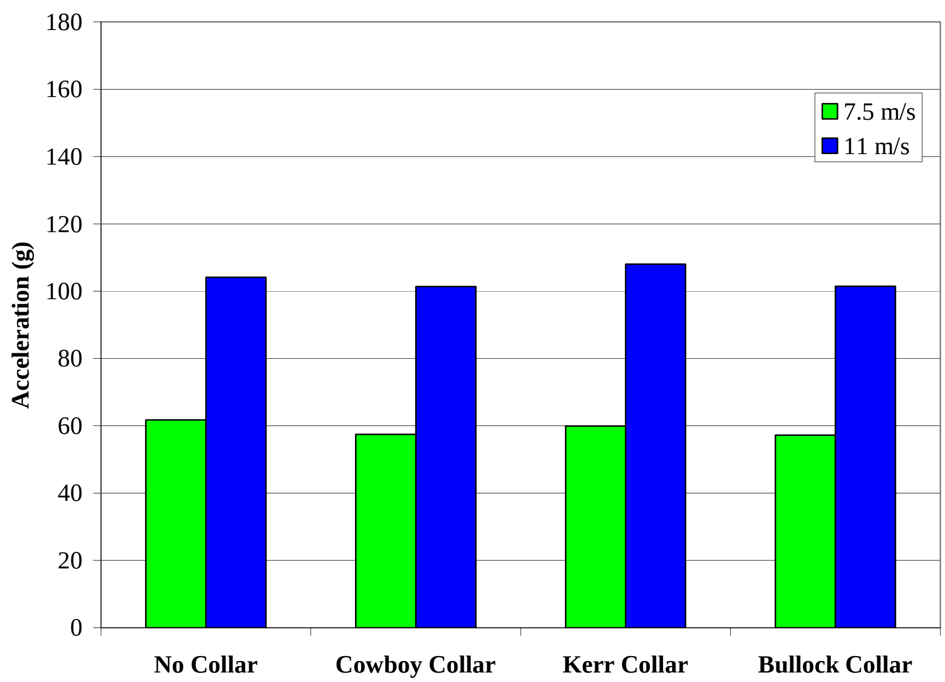

Figure A 14: Side impact location, raised shoulder pads, peak head acceleration

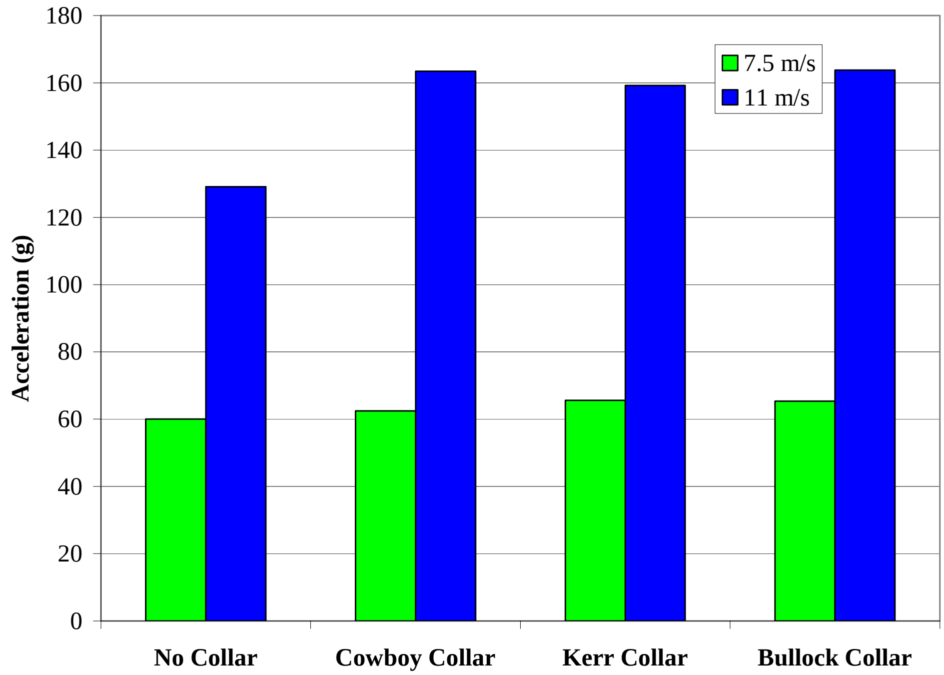

Figure A 15: Front impact location, peak head acceleration

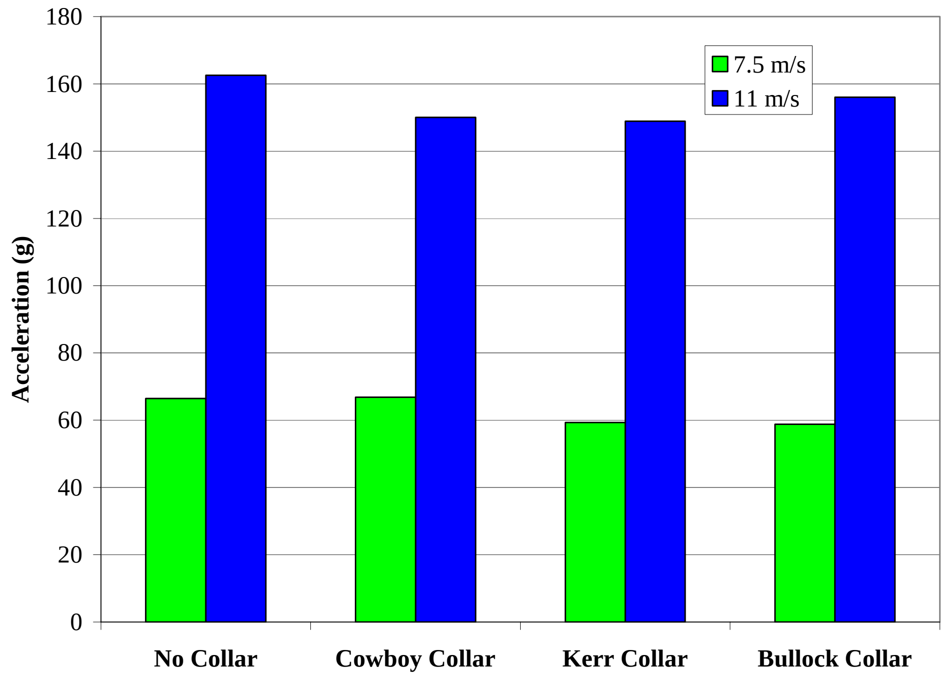

Figure A 16: Front impact location, raised shoulder pads, peak head acceleration

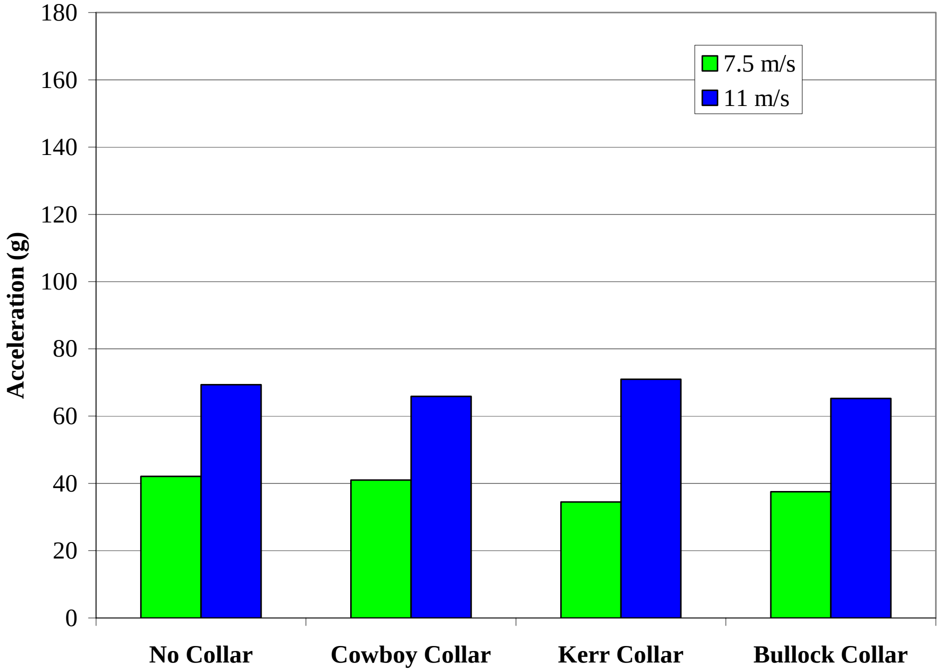

Figure A 17: Axial loading impact location, peak head acceleration

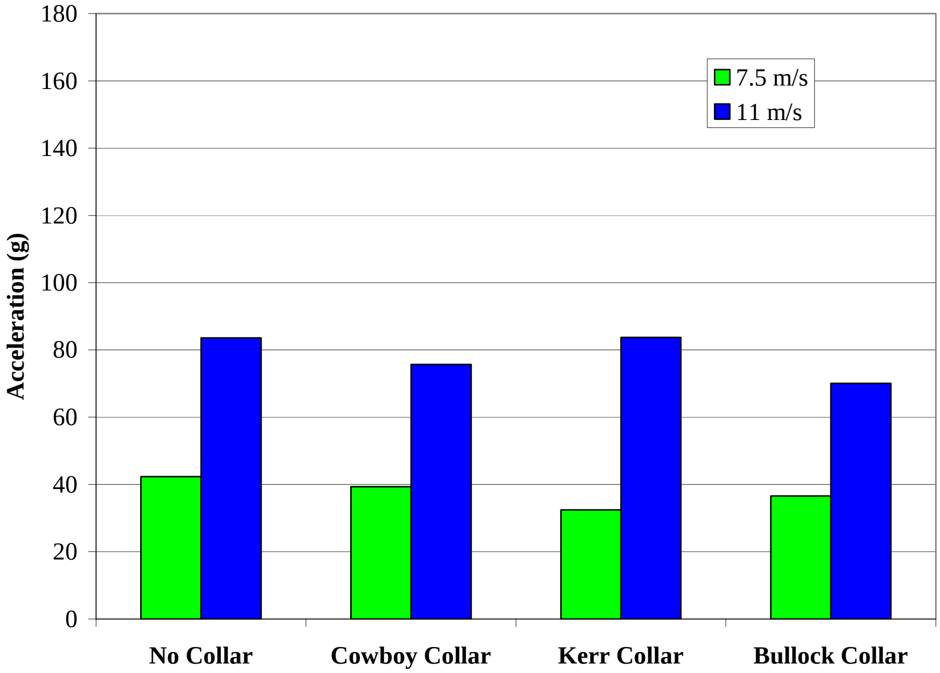

Figure A 18: Axial loading impact location, raised shoulder pads, peak head acceleration

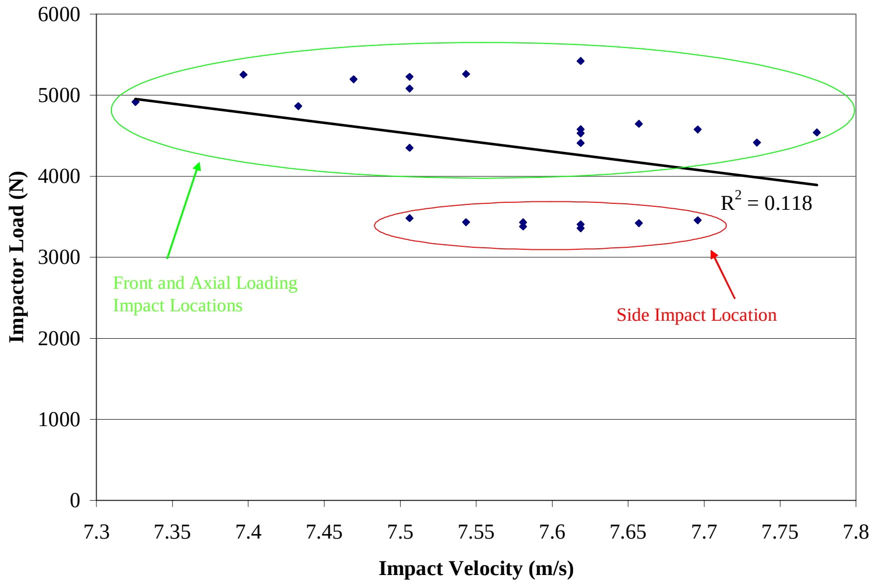

Figure A 19: Impact velocity versus impactor load, low impact velocity

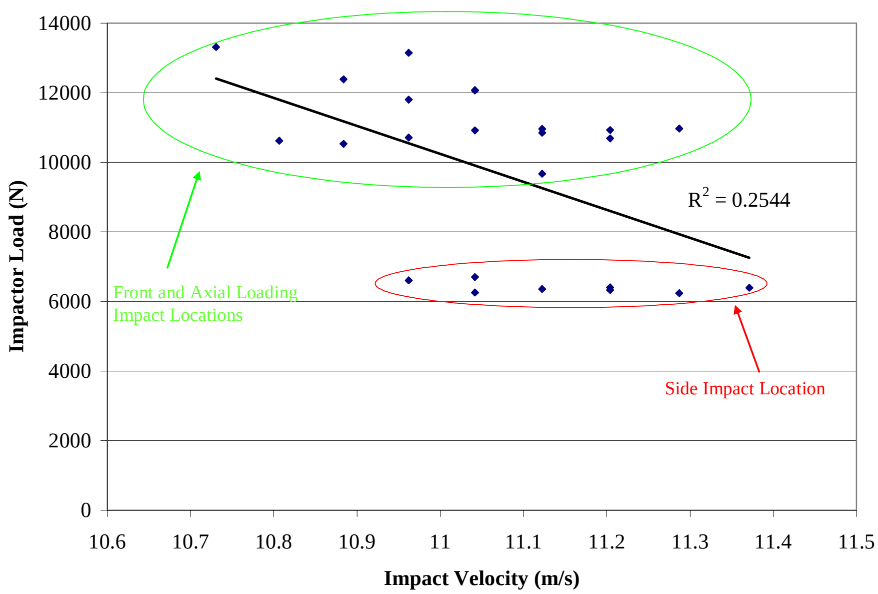

Figure A 20: Impact velocity versus impactor load, high impact velocity

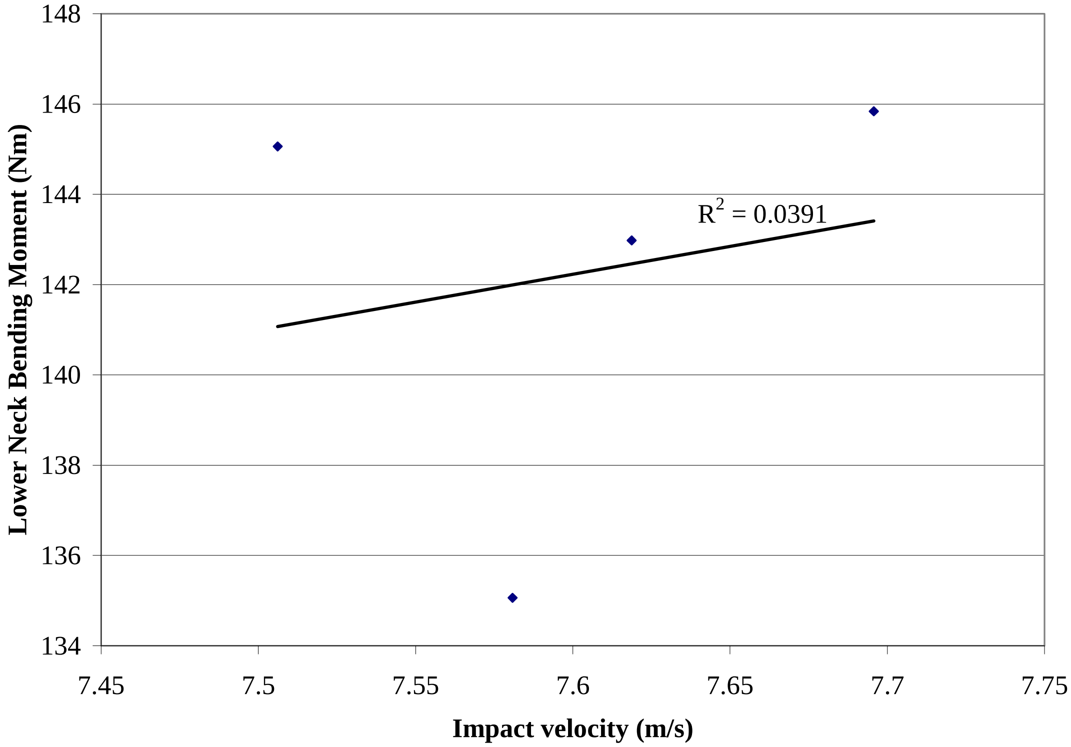

Figure A 21: Impact velocity versus lower neck bending moment, side impact location, low impact velocity

APPENDIX B: Test setup

Figure B 1: Side impact location (front view)

Figure B 2: Side impact location (oblique view)

Figure B 3: Side impact location (front view)

Figure B 4: Front impact location (oblique view)

Figure B 5: Axial loading impact location (front view)

Figure B 6: Axial loading impact location (oblique view)

Figure B 7: Cowboy Collar in raised position

Figure B 8: Cowboy Collar in raised position (closeup)

Figure B 9: Kerr collar in normal position

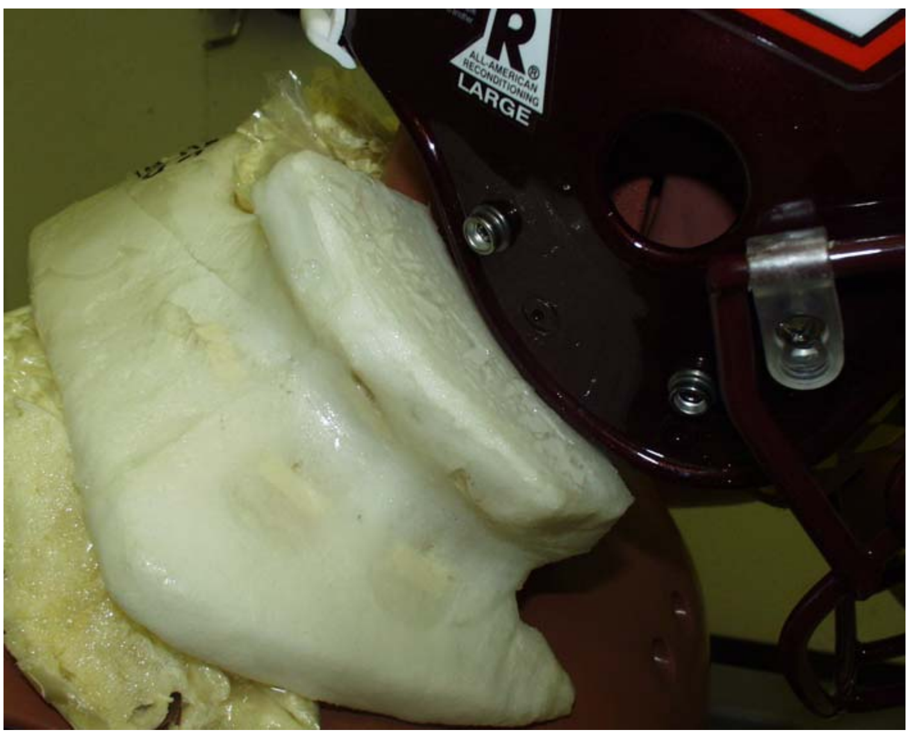

Figure B 10: Kerr collar in raised position