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Biomechanical Analysis of Neck Collars Used in Football

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ABSTRACT

The objective of this study was to perform a dynamic biomechanical analysis of neck collars in order to determine their effect on head and neck loading. A total of 48 tests were performed comparing the Cowboy Collar, Bullock Collar, and the Kerr Collar to control tests at two speeds (5 m/s and 7 m/s), three impact locations (front, top, and side of the helmet), and two shoulder pad positions (normal and raised). A 50th percentile male Hybrid III dummy was equipped with a helmet, shoulder pads, and the neck collars. The dummy was instrumented with tri-axial accelerometers at the center of gravity of the head. Angular rate sensors were used in the head and chest. In addition, both the upper and lower neck were instrumented with six-axis load cells. The helmet was struck with a NOCSAE style pneumatic linear impactor to provoke rotation of the head and neck. In the top impact location, it was found that the Kerr Collar and Bullock Collar reduced head accelerations and force transmission through the neck with the Kerr Collar producing greater reductions in force transmission. In the front impact location, all the collars reduced lower neck moment. The Kerr Collar was also capable of reducing the lower neck force and upper neck moment. In the side impact location, the Kerr Collar substantially reduced lower neck moment. These reductions in loads correlate with the degree to which each collar restricted the motion of the head and neck. By restricting the range of motion of the neck and redistributing load to the shoulders, neck loads were effectively lowered.

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1 INTRODUCTION

Neck injuries in football can vary from the rare catastrophic event, to the much more frequent but less severe neck stinger. Specifically, brachial plexopathy, known as a stinger or burner, is a common injury in competitive football. Studies have shown lifetime injury incidences from 49% to 65% in college football (Clancy, 1977; Sallis, 1992). Many players will wear neck collars to prevent such injuries. These collar designs are based off empirical data and few experiments have been conducted to quantify their effectiveness.

1.1 INJURY MECHANISMS

A stinger is most likely caused by damaging 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. Usually, these symptoms 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).

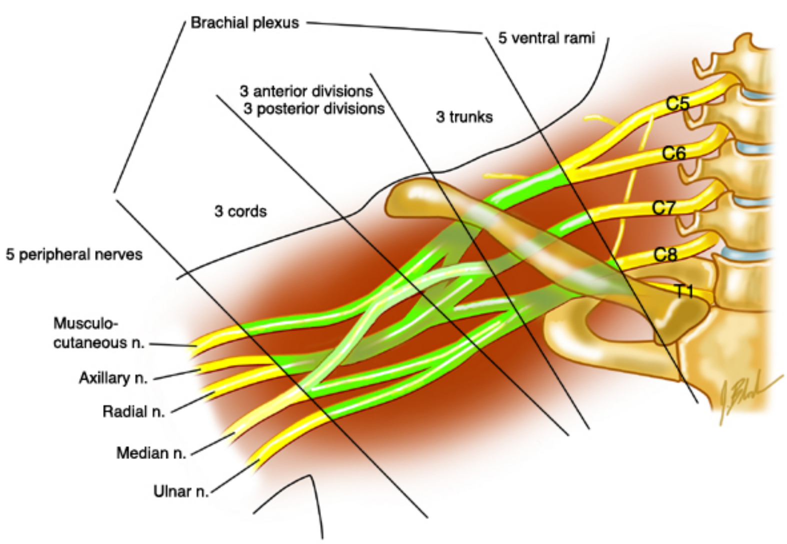

Figure 1: Brachial plexus (photo: http://www.backpain-guide.com)

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.

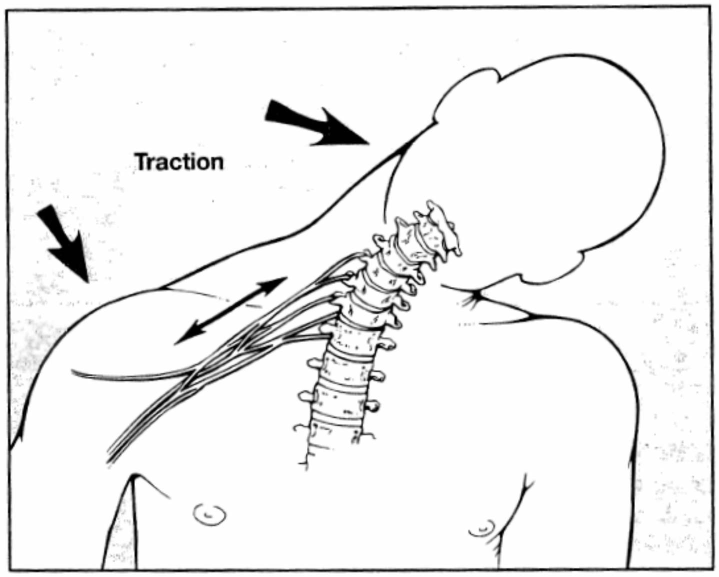

Figure 2: Stinger injury mechanism, traction (Sallis, 1992)

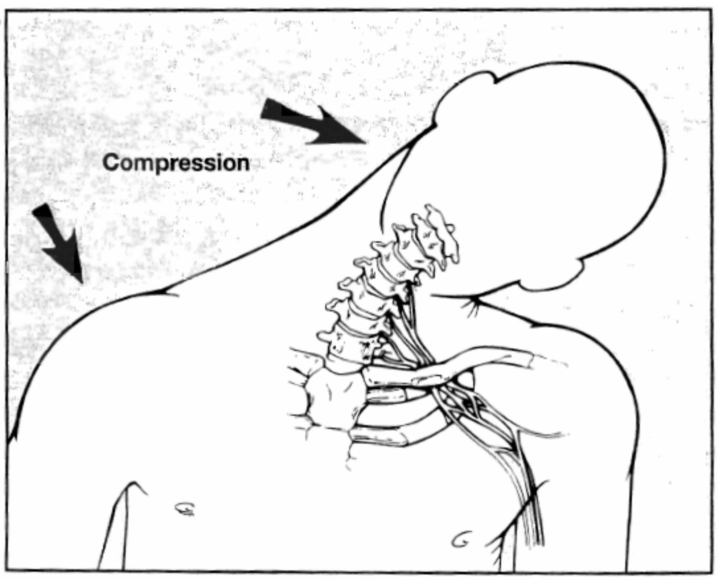

Figure 3: Stinger injury mechanism, compression (Sallis, 1992)

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 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.

1.2 PREVIOUS RESEARCH

The collars that are worn by football players to prevent this injury were most often designed and put into use without biomechanical testing. 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, 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).

1.3 OBJECTIVE

The objective of this study was to perform a biomechanical analysis of neck collars through dynamic testing. Dynamic tests are useful because they more closely simulate the type of impacts that are actually seen on the playing field.

2 METHODOLOGY

An instrumented 50^{th} percentile male Hybrid III test dummy was used to assess the effectiveness of various neck protection devices used in football. The dummy was suited with a set of Douglas CP25 shoulder pads and a Riddell VSR4 helmet for all tests. A NOCSAE style pneumatic linear impactor was used to strike the helmet in various locations at several speeds. Control tests were performed in which the dummy was equipped with shoulder pads, a helmet, and no neck collar. Control tests were done for each impact speed and location. Following the control tests, the same tests were repeated with the dummy wearing the various neck collars. The effectiveness of the collars could be determined by comparing the data from the control tests with the data from the collar tests.

2.1 NECK COLLARS

Three different neck collars were evaluated in this set of testing. The first collar tested was the Cowboy Collar, manufactured by McDavid. Its design consists of a molded polyurethane foam collar that gets laced into the shoulder pads. The Cowboy Collar is designed to limit extension of the neck and has the least side protection of any of the collars tested. The second collar tested was the Bullock Collar, which was designed by the Virginia Tech head team physician, Richard Bullock. Its design consists of a high-density foam collar with a rigid plastic insert that is strapped to the shoulder pads. The Bullock Collar is designed to prevent hyperextension of the neck and provides some support on the sides of the neck. The last collar tested was the Kerr Collar, a prototype collar designed by Patrick Kerr. Its design consists of a molded synthetic collar that rests on the shoulders and gets laced into the shoulder pads. The Kerr Collar is designed so that the base of the helmet contacts the collar, restricting motion. Figure 4 is a graphic displaying the three different collars tested (from left to right: Cowboy Collar, Bullock Collar, Kerr Collar).

Figure 4: Cowboy, Bullock, and Kerr Collars (shown left to right)

2.2 TEST PARAMETERS

Impact velocities were chosen so that they would encompass the impacts typically seen in football games. The impacting speeds used were 5 $^{m}/_{s}$, and 7 $^{m}/_{s}$. The 5 $^{m}/_{s}$ test would represent a lower speed impact, while the 7 $^{m}/_{s}$ tests would represent a more severe impact. Impact locations were also chosen based on impacts seen in football games. The locations impacted were the side, front, and top of the helmet (Figure 5). The side location was in the center of the side of the helmet so that impacts there would promote pure lateral bending of the neck. The front location was just above where the facemask meets the helmet shell. Impacts at this location forced the neck of the dummy into extension. The top location was impacted at the top of the helmet, above the facemask. Impacting the top allowed axial compression of the neck to be evaluated. In an effort to simulate actual football impacts and to account for the various ways a player may get hit, different shoulder pad positions were also tested. This involved testing the shoulder pads in a normal and raised position. In order to raise the shoulder pads, shoulder implants were made for the dummy using expanding polyurethane foam. These implants were secured on the shoulders of the dummy for the raised shoulder pad tests. The different combinations of these variables [4 collars x 2 speeds x 3 locations x 2 shoulder pad positions] resulted in 48 tests.

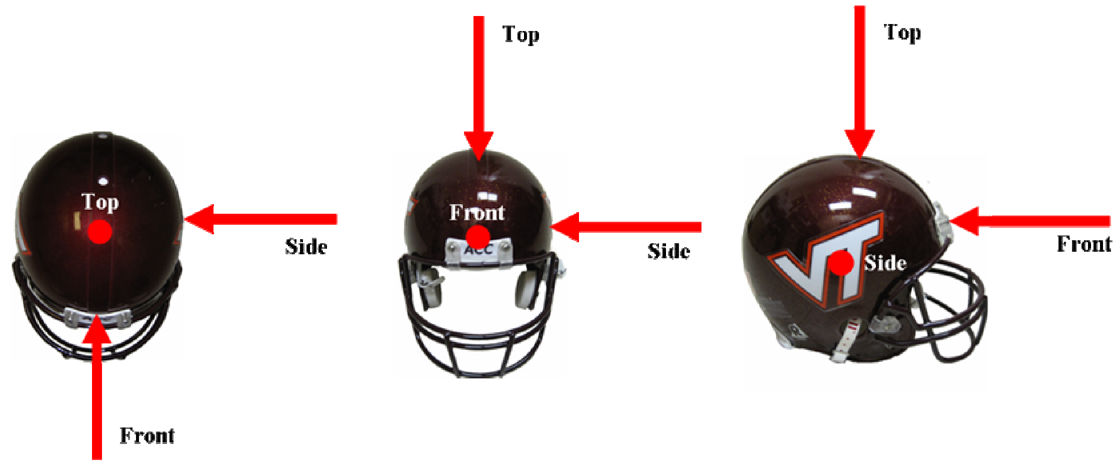

Figure 5: Top, Front, and Side Impact Locations

2.3 Instrumentation and Equipment

The dummy was fitted with three single-axis orthogonally mounted accelerometers and a tri-axial angular rate sensor in the center of gravity of the head (Figure 6 and Table 1). The chest of the dummy was also fitted with an angular rate sensor. The dummy was instrumented with angular rate sensors only during the 5 m/s tests. The head and neck's range of motion could be calculated from the angular rate data using a technique described by Hall (1998). High speed video was used to confirm the calculated angles. The neck was instrumented with upper and lower neck load cells which provided forces and moments for each axis. The impactor was instrumented with a load cell on the impactor arm to measure impact force. The impactor arm was also instrumented with an accelerometer. A light gate was used to measure the velocity of the impactor arm as it contacted the dummy. All instrumentation was sampled at 10,000 Hz and processed in accordance with SAE J211. A digital high speed color camera recorded each test at 1000 frames per second.

Table 1: Hybrid III and Impactor Instrumentation

Measurement		Instrument	Model Number
Head Acceleration	X	Accelerometer	Endevco B40351
	Y	Accelerometer	Endevco B40234
	Z	Accelerometer	Endevco B40740
Chest Angular Rate	X	Angular Rate Sensor	ARS-129
	Y	Angular Rate Sensor	ARS-132
	Z	Angular Rate Sensor	ARS-131
Lower Neck Force and Moments		Load Cell	LC-242
Upper Neck Force and Moments		Load Cell	LC-592
Impactor Acceleration		Accelerometer	Endevco B40592
Impactor Force		Load Cell	LC-91
Head Angular Rates		Angular Rate Sensor	IES-3103

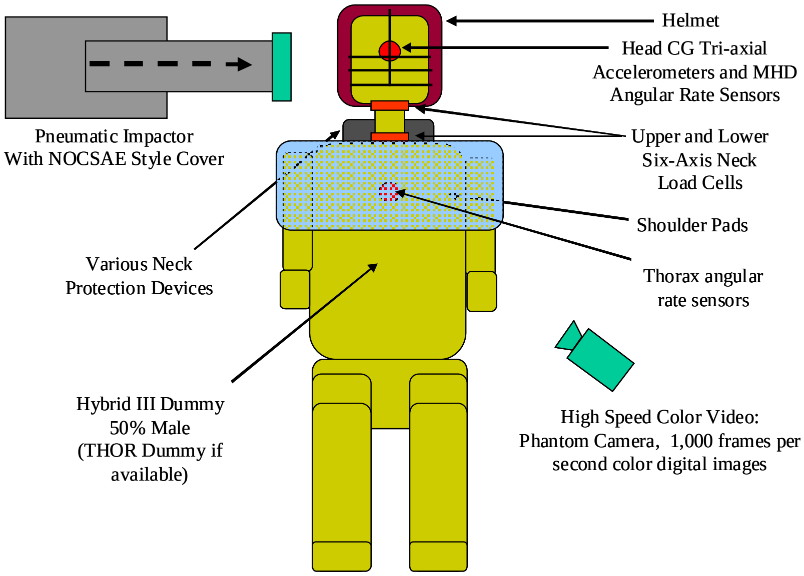

Figure 6: Experimental Setup and Instrumentation Schematic

The impacts were performed with a pneumatic linear impactor (Figure 7). When triggered, the solenoid opens the butterfly valve and sends a blast of compressed air into the shaft. This results in the piston accelerating out of the shaft and pushing the impactor arm into the dummy. The impactor arm weighs 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, 2003). This impact surface is elastic and has a minimal rest period between impacts. In addition, its impact properties do not change over the course of testing.

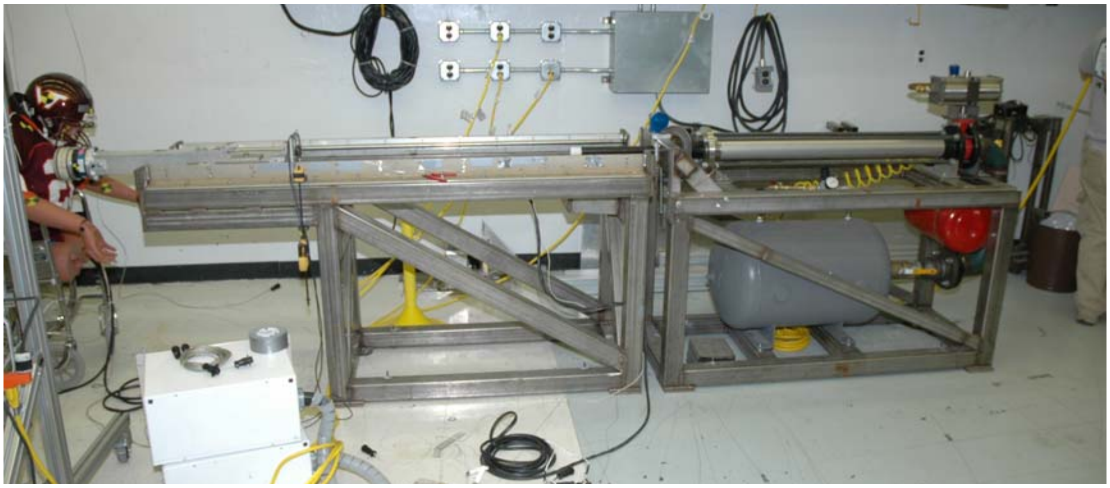

Figure 7: Pneumatic Linear Impactor

To ensure that the positioning of the helmet on the dummy was consistent, a helmet positioning tool was used. The tool uses landmarks on the helmet and dummy's face to position the helmet. In addition, the position of the dummy relative to the impactor was precisely controlled. Targets on the helmet were aligned with the center of the impactor.

3 RESULTS

This section is partitioned by impact location. Results will be presented for the different areas of interest at each location. Figure 8 displays the sign convention used that will be used in this paper. The positive x-axis runs out of the face, the positive y-axis runs out of the right ear, and the positive z-axis runs out of the bottom of the head.

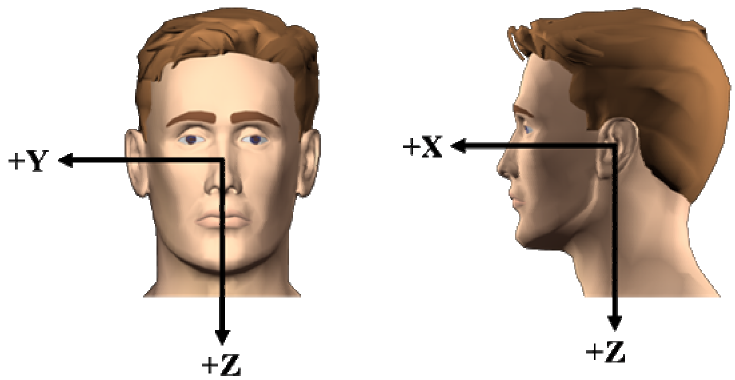

Figure 8: Coordinate System Used for Data Analysis

3.1 TOP IMPACT LOCATION

An impact to the top of the helmet promotes axial compression of the neck. The most relevant data in a top impact test are the resultant head acceleration, force transmission reduction in the lower neck, and force transmission reduction in the upper neck. Force transmission reduction in the upper and lower neck was calculated as a percentage. This percentage represents the fraction of the impactor load not experienced in the upper and lower neck. For example, if the impactor force is 1000 N and the lower neck force is 700 N, then there is a 300 N reduction, or a 30 % reduction in the lower neck. Table 2 displays the peak values for each collar in the normal shoulder pad position when tested at each velocity. Table 3 presents the results of the top impact, raised shoulder pads configuration.

Table 2: Peak Values for Top Impact, Normal Shoulder Pad Configuration

Top Impact						
Normal Shoulder Pad Position		Neck Collar				Units
		None	Cowboy	Bullock	Kerr	
5 m/s	Test ID	top1	top2	top3	top4	
	Actual Velocity	4.90	4.97	4.93	4.95	m/s
	Resultant Head Acceleration	31	31	28	24	G
	Impactor Force	3533	3535	3630	3751	N
	Lower Neck Force (Fz)	3550	3553	3181	3197	N
	Lower Neck % Force Reduction	0	0	10	15	%
	Upper Neck Force (Fz)	4210	4211	3751	3651	N
	Upper Neck % Force Reduction	0	0	0	3	%
7 m/s	Test ID	top5	top6	top7	top8	
	Actual Velocity	6.44	6.41	6.49	6.25	m/s
	Resultant Head Acceleration	42	41	34	34	G
	Impactor Force	5203	5378	5260	6821	N
	Lower Neck Force (Fz)	4496	4410	4039	4677	N
	Lower Neck % Force Reduction	14	18	23	31	%
	Upper Neck Force (Fz)	5334	5240	4718	5609	N
	Upper Neck % Force Reduction	0	3	10	18	%

Table 3: Peak Values for Top Impact, Raised Shoulder Pad Configuration

Top Impact						
Raised Shoulder Pad Position		Neck Collar				Units
		None	Cowboy	Bullock	Kerr	
5 m/s	Test ID	top9	top10	top11	top12	
	Actual Velocity	4.92	4.93	4.92	4.87	m/s
	Resultant Head Acceleration	29	30	28	22	G
	Impactor Force	3574	3554	3451	3546	N
	Lower Neck Force (Fz)	3205	3301	3387	2331	N
	Lower Neck % Force Reduction	10	7	2	34	%
	Upper Neck Force (Fz)	3781	3875	4005	2761	N
	Upper Neck % Force Reduction	0	0	0	22	%
7 m/s	Test ID	top13	top14	top15	top16	
	Actual Velocity	6.58	6.41	6.36	6.25	m/s
	Resultant Head Acceleration	41	39	32	32	G
	Impactor Force	5006	4823	5196	6226	N
	Lower Neck Force (Fz)	4262	4315	3729	3525	N
	Lower Neck % Force Reduction	15	11	28	43	%
	Upper Neck Force (Fz)	5034	5079	4380	4322	N
	Upper Neck % Force Reduction	0	0	16	31	%

The Kerr Collar and the Bullock Collar consistently lowered head accelerations resulting from a top impact. However, the Kerr Collar produced greater head accelerations reductions (Figure 9). The Kerr Collar also reduced force transmission through the neck. In every testing scenario, the Kerr Collar reduced the most force. The Bullock Collar also performed well, but only in the higher velocity tests. Figure 10 and Figure 11 show these force reductions graphically.

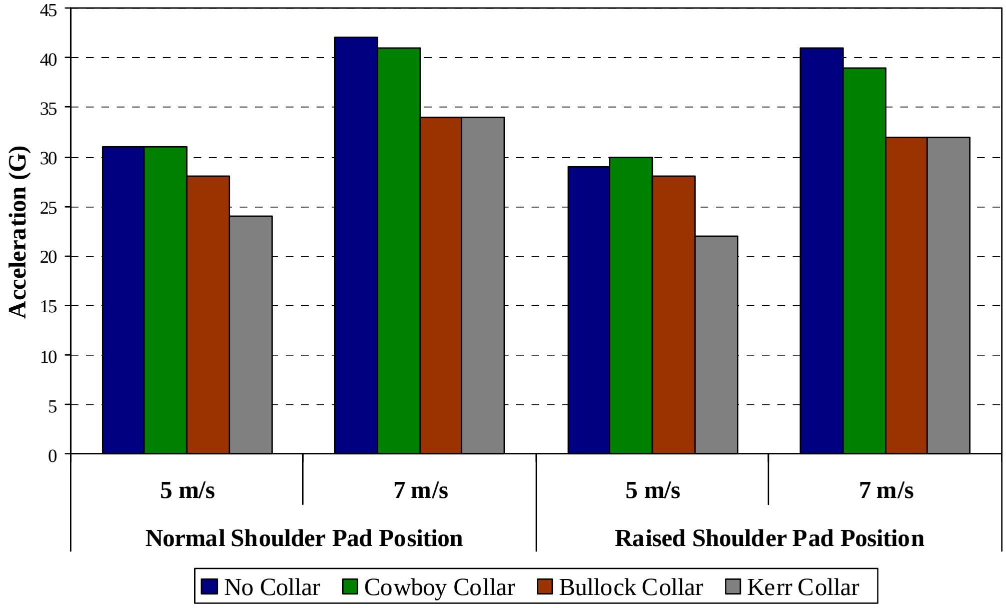

Figure 9: Resultant Head Acceleration Resulting From Top Impact

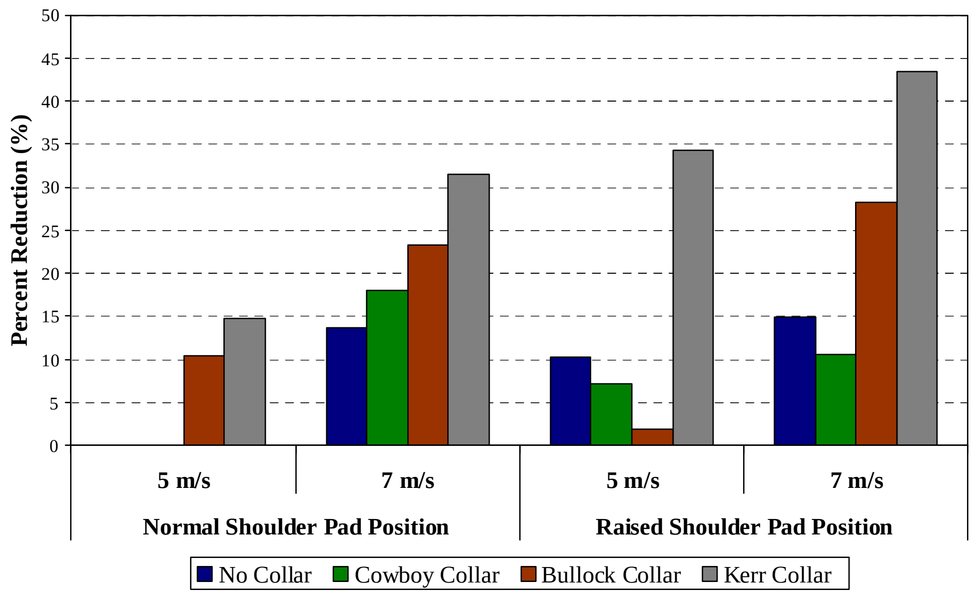

Figure 10: Top Impact - Lower Neck Force Transmission Reduction

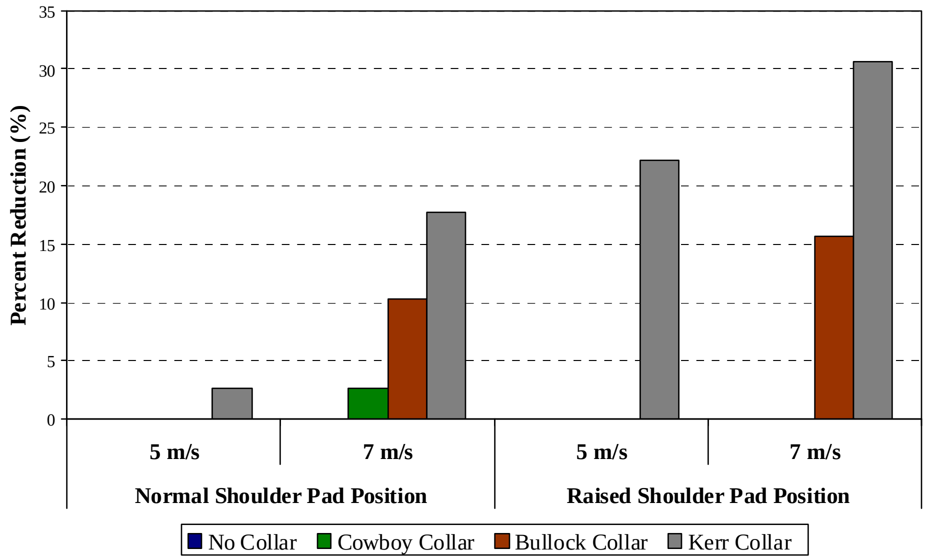

Figure 11: Top Impact - Upper Neck Force Transmission Reduction

3.2 FRONT IMPACT LOCATION

A front impact promotes extension of the neck. The most relevant data in a front impact test are the resultant head acceleration, upper and lower neck forces along the x-axis, and upper and lower neck moment about the y-axis. Table 4 displays the peak values for each collar in the normal shoulder pad position when tested at each velocity. Table 5 presents the results of the side impact, raised shoulder pads configuration.

Table 4: Peak Values for Front Impact, Normal Shoulder Pad Configuration

Front Impact						
Normal Shoulder Pad Position		Neck Collar				Units
		None	Cowboy	Bullock	Kerr	
5 m/s	Test ID	front1	front2	front3	front4	
	Actual Velocity	4.90	4.97	4.92	4.92	m/s
	Impactor Force	3081	3043	2772	2894	N
	Resultant Head Acceleration	49	48	47	43	G
	Lower Neck Force (Fx)	611	621	567	387	N
	Lower Neck Moment (My)	86	82	77	89	N*m
	Upper Neck Force (Fx)	613	594	559	558	N
	Upper Neck Moment (My)	51	49	54	35	N*m
	Range of Motion	25	23	19	15	deg
7 m/s	Test ID	front5	front6	front7	front8	
	Actual Velocity	6.52	6.52	6.55	6.25	m/s
	Impactor Force	4410	4530	4579	4870	N
	Resultant Head Acceleration	60	62	65	70	G
	Lower Neck Force (Fx)	676	650	707	533	N
	Lower Neck Moment (My)	133	127	117	139	N*m
	Upper Neck Force (Fx)	704	710	741	773	N
	Upper Neck Moment (My)	64	67	73	41	N*m

Table 5: Peak Values for Front Impact, Raised Shoulder Pad Configuration

Front Impact						
Raised Shoulder Pad Position		Neck Collar				Units
		None	Cowboy	Bullock	Kerr	
5 m/s	Test ID	front9	front10	front11	front12	
	Actual Velocity	4.89	4.98	4.85	4.90	m/s
	Impactor Force	2828	2911	2886	2811	N
	Resultant Head Acceleration	49	46	49	42	G
	Lower Neck Force (Fx)	602	623	560	415	N
	Lower Neck Moment (My)	78	72	87	54	N*m
	Upper Neck Force (Fx)	566	577	583	505	N
	Upper Neck Moment (My)	49	49	50	25	N*m
	Range of Motion	19	16	17	12	deg
7 m/s	Test ID	front13	front14	front15	front16	
	Actual Velocity	6.61	6.58	6.52	6.25	m/s
	Impactor Force	4577	4416	4351	4580	N
	Resultant Head Acceleration	66	67	59	66	G
	Lower Neck Force (Fx)	742	673	589	625	N
	Lower Neck Moment (My)	125	105	123	90	N*m
	Upper Neck Force (Fx)	747	661	674	726	N
	Upper Neck Moment (My)	73	60	56	32	N*m

No collar produced any reductions in head acceleration with this impact location. Any difference in head acceleration fell within the repeatability variations of the tests. The Kerr Collar reduced lower neck force more than any other collar. This can be seen in Figure 12. The Cowboy Collar and Bullock Collar produced some reduction in lower neck moment. The Kerr Collar produced lower neck moment reductions when the shoulder pads were in the raised position. Figure 13 displays this. No collar was capable of substantially reducing the upper neck force, but it is worth noting that the collars seemed to perform better when in the raised position. Upper neck moment was only consistently reduced by the Kerr Collar. This reduction is shown in Figure 14. Although all the collars restricted the head and neck's range of motion, the Kerr Collar allowed the least amount of movement.

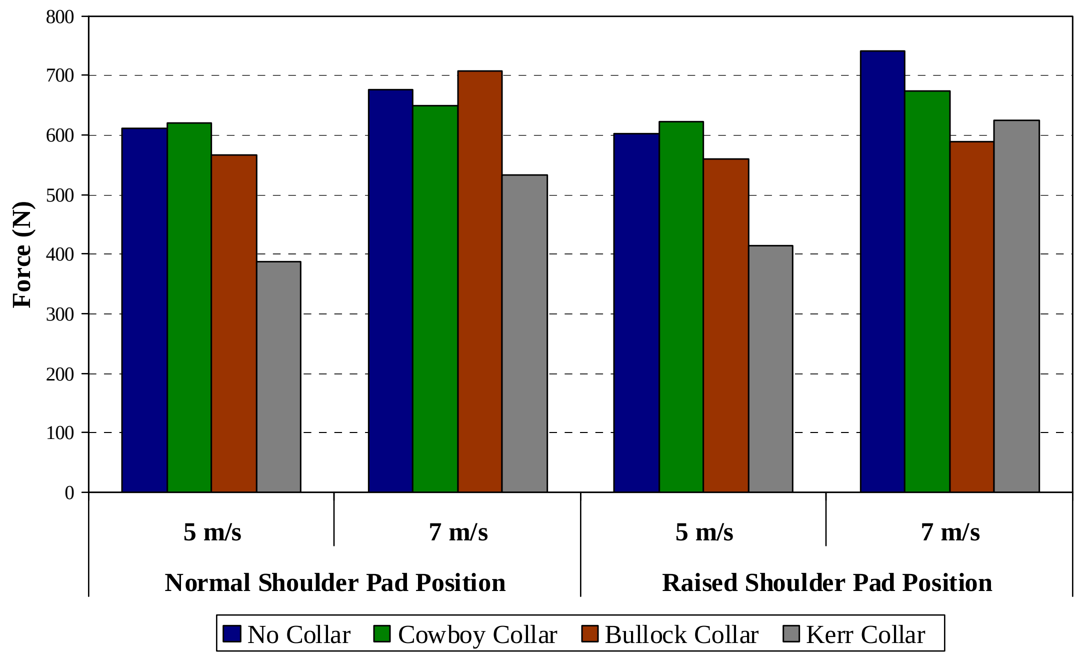

Figure 12: Lower Neck Force (Fx) Resulting From Front Impact

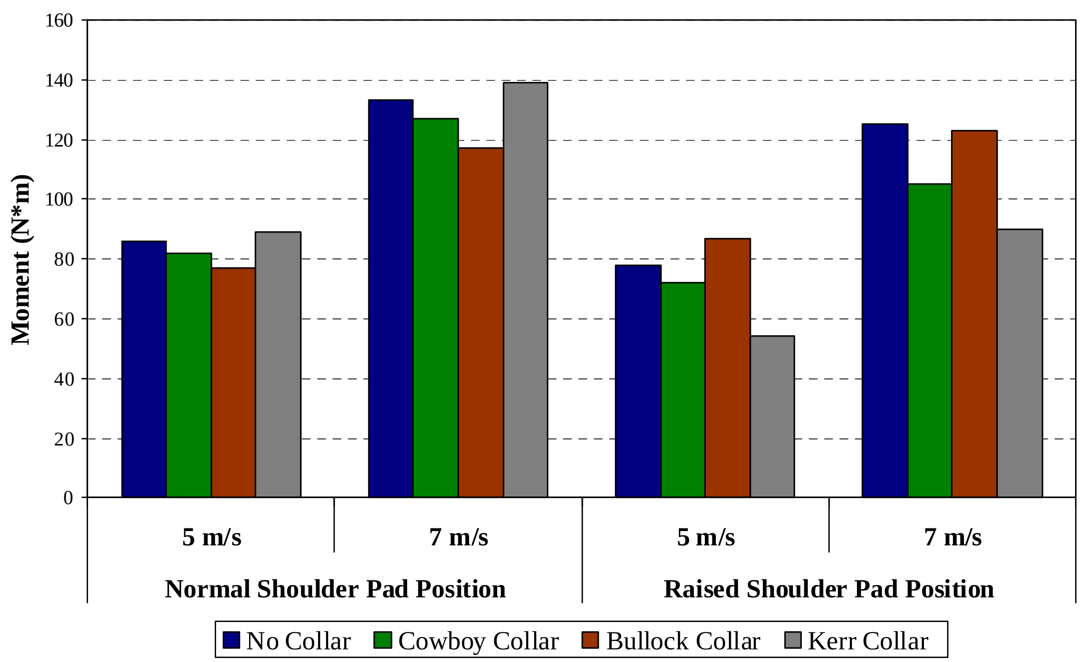

Figure 13: Lower Neck Moment (My) Resulting From Front Impact

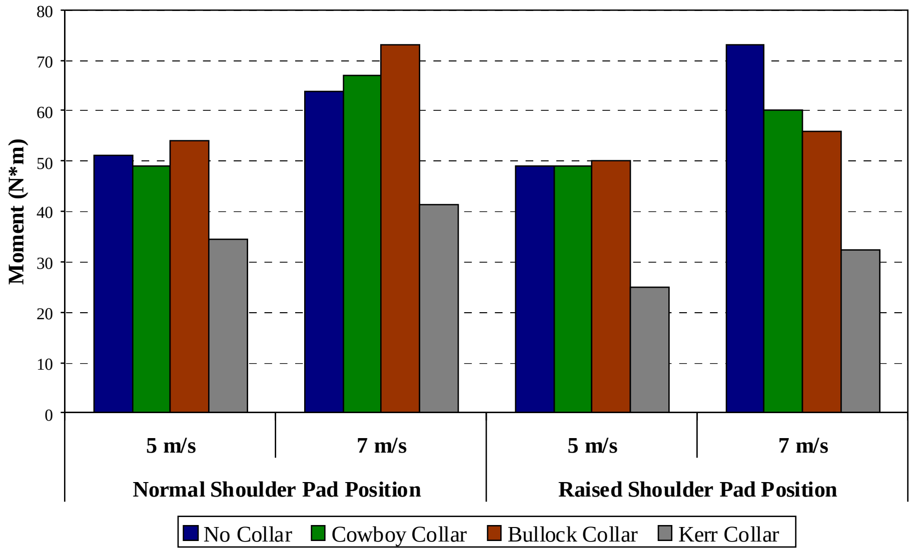

Figure 14: Upper Neck Moment (My) Resulting From Front Impact

3.3 SIDE IMPACT LOCATION

A side impact promotes lateral bending of the neck. The most relevant data in a side impact test are the resultant head acceleration, upper and lower neck forces along the y-axis, and upper and lower neck moment about the x-axis. Table 6 displays the peak values for each collar in the normal shoulder pad position when tested at each velocity. Table 7 presents the results of the side impact, raised shoulder pads configuration.

Table 6: Peak Values for Side Impact, Normal Shoulder Pad Configuration

Side Impact						
Normal Shoulder Pad Position		Neck Collar				Units
		None	Cowboy	Bullock	Kerr	
5 m/s	Test ID	side1	side2	side3	side4	
	Actual Velocity	4.90	4.93	4.92	4.95	m/s
	Impactor Force	2781	2893	2843	3040	N
	Resultant Head Acceleration	54	64	61	61	G
	Lower Neck Force (Fy)	452	442	430	398	N
	Lower Neck Moment (Mx)	127	117	112	110	N*m
	Upper Neck Force (Fy)	530	526	501	554	N
	Upper Neck Moment (Mx)	35	34	30	35	N*m
	Range of Motion	39	37	37	25	deg
7 m/s	Test ID	side5	side6	side7	side8	
	Actual Velocity	6.85	6.82	6.84	6.85	m/s
	Impactor Force	3651	3968	3732	3646	N
	Resultant Head Acceleration	75	79	78	75	G
	Lower Neck Force (Fy)	607	626	537	557	N
	Lower Neck Moment (Mx)	154	153	157	136	N*m
	Upper Neck Force (Fy)	621	676	579	619	N
	Upper Neck Moment (Mx)	57	61	59	53	N*m

Table 7: Peak Values for Side Impact, Raised Shoulder Pad Configuration

Side Impact						
Raised Shoulder Pad Position		Neck Collar				Units
		None	Cowboy	Bullock	Kerr	
5 m/s	Test ID	side9	side10	side11	side12	
	Actual Velocity	4.89	4.98	4.85	4.92	m/s
	Impactor Force	2611	2720	2494	2797	N
	Resultant Head Acceleration	54	54	51	57	G
	Lower Neck Force (Fy)	419	421	427	398	N
	Lower Neck Moment (Mx)	112	108	112	91	N*m
	Upper Neck Force (Fy)	449	430	471	475	N
	Upper Neck Moment (Mx)	31	35	30	36	N*m
	Range of Motion	33	34	33	17	deg
7 m/s	Test ID	side13	side14	side15	side16	
	Actual Velocity	6.81	6.82	6.85	6.88	m/s
	Impactor Force	3830	3796	3860	3549	N
	Resultant Head Acceleration	73	74	80	69	G
	Lower Neck Force (Fy)	584	517	631	530	N
	Lower Neck Moment (Mx)	149	132	146	118	N*m
	Upper Neck Force (Fy)	572	491	650	406	N
	Upper Neck Moment (Mx)	55	47	54	58	N*m

No collar was capable of reducing head accelerations with this impact location. Any difference in head acceleration fell within the repeatability variations of the tests. In addition, no collar substantially reduced the lower or upper neck forces resulting from the side impact. The Kerr Collar reduced the lower neck moment more than the other collars tested. Figure 15 represents this graphically. No collar was capable of reducing the upper neck moment. The Kerr Collar was the only collar that reduced lateral bending of the neck.

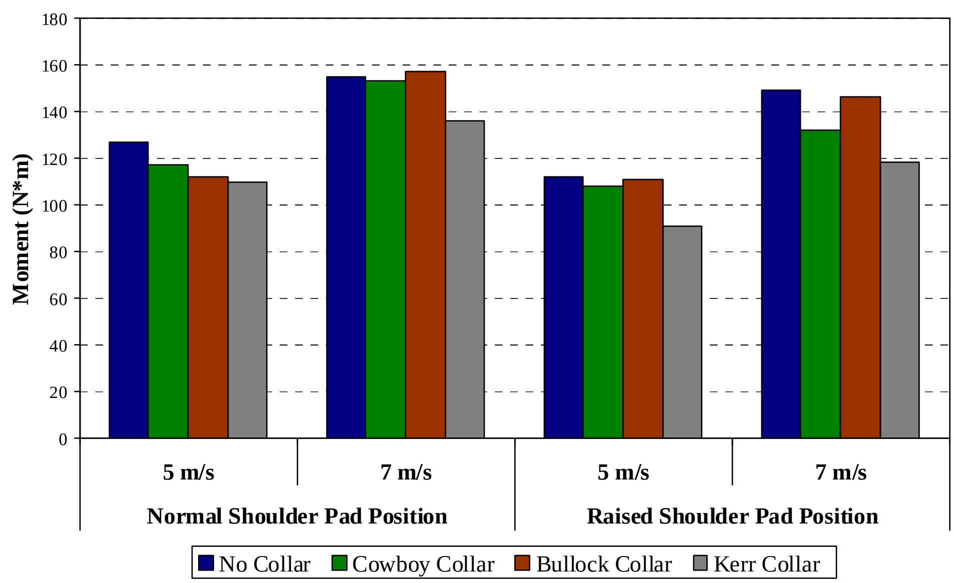

Figure 15: Lower Neck Moment (My) Resulting From Side Impact

4 LIMITATIONS

4.1 BIOFIDELITY

The Hybrid III test dummy's neck has limited biofidelity. The anatomy of the cervical spine is complicated, specifically in how the vertebrae articulate with each other. In addition, the neck has small muscle groups that provide support and strength. These are all characteristics the Hybrid III neck lacks. The Hybrid III neck models the neck as butyl rubber segmented by aluminum discs. It also has a steel cable running through the center of the neck to control extension/flexion. While the Hybrid-III neck responds similarly to a human's in a frontal crash test, it cannot perfectly replicate the complicated kinematics that occur during such impacts, and thus is an imperfect model. Figure 16 compares the cervical spine to the Hybrid III neck.

Figure 16: Anterior and Lateral Views of the Cervical Spine (left) and Hybrid III Neck (right)

Even though there are biofidelity issues with the Hybrid III test dummy, it is the best available surrogate for a human. It has become the standard for the automotive industry when testing to predict injuries in crash tests. The Hybrid III has also been used for various other applications, including football testing. Pellman (2003) reconstructed concussive football impacts with Hybrid III dummies using multiple angles of game video. The Hybrid III dummy may not be perfectly biofidelic, but it is commonly accepted as a human surrogate and these types of tests have been performed in the past (Duma, 2005). Moreover, the effect of the limited biofidelity is reduced by examining trends in the data, which is the goal of this report. It is also important to note that given the dummy's stiff neck, caution should be used before comparing the output data to published injury thresholds.

4.2 EXPERIMENTAL VARIATION

Since only one test was done for each collar/shoulder pad/speed configuration, repeatability tests were conducted in order to determine which differences in performance could be considered significant. These tests proved there was some variability inherent in the test setup. However, the variance was an acceptable amount and some data channels proved more sensitive than others. Five consecutive tests in the control configuration were performed to assess the repeatability of this experiment. Percent deviations were determined by dividing the standard deviation by the average peak value for each data channel. Head acceleration was the most sensitive data channel with 6.7% deviation. Upper and lower neck moments had 1.6% and 2.9% variation between the tests, respectively. Upper and lower neck forces varied by 4.5% and 3.9% respectively. Although these values may seem a little high, the neck collars' performances differed enough for these values to be used to determine significant differences between collars. In addition, these curves followed the same data-traces through time, which allowed for comparison between characteristics of the data throughout time. Figure 17 displays the lower neck moment curves for a set of five side impact tests with the same configuration.

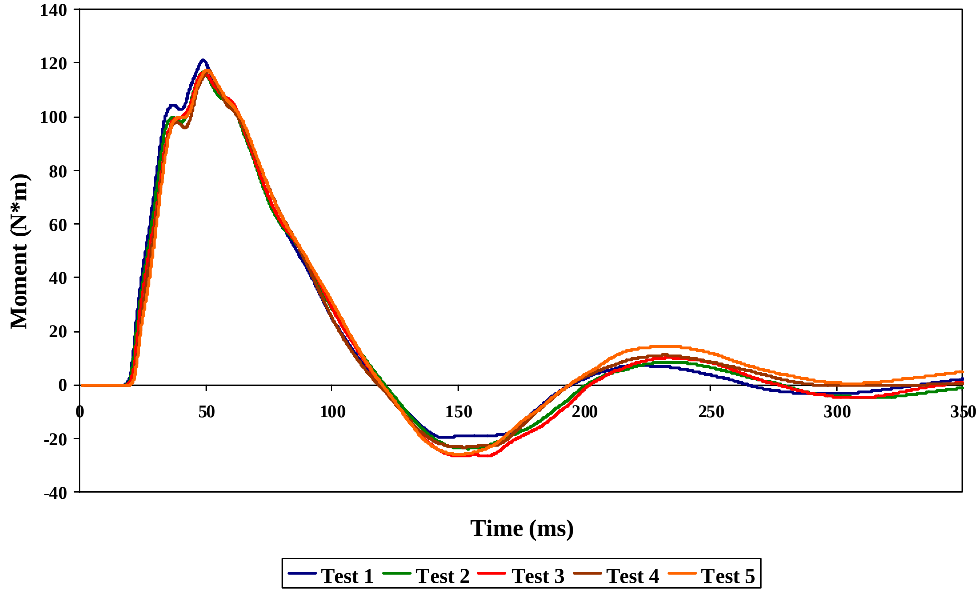

Figure 17: Lower Neck Moment (Mx) Curves for Repeatability Tests

5 DISCUSSION

In a top impact, the Kerr Collar provided the most protection. The Bullock Collar provided some protection, while the Cowboy Collar did not protect the dummy from experiencing high neck loads. The Kerr Collar reduced the head acceleration and force transmission due to its unique design. The Kerr Collar is designed to contact the base of the helmet during an impact. In a top impact, this redirects some of the load to the shoulders, on which the collar rests. The stiffness of the collar prevents the neck from further compression. Neither the Bullock Collar nor Cowboy Collar prevents the neck from compressing. However, the Bullock Collar was capable of reducing a small portion of the load in some configurations. This is most likely due to the back of the helmet contacting the collar.

The Kerr Collar also provided the most protection during an impact to the front of the helmet. It reduced upper neck moment and lower neck force in all configurations. The Kerr Collar also reduced the lower neck moment, but only in the raised configuration. Upon inspection of the high speed video, the collar restricts the range of motion of the head and neck by contacting the base of the helmet during the impact. This contact between the helmet and collar is responsible for the lower loads.

The Kerr Collar typically performed better in the raised position because it contacts the collar sooner and restricts more motion. This is true for any of the collars in the raised position. The Cowboy Collar and Bullock Collar also provided protection for the dummy throughout the front impacts. The reductions of loads were not as large and consistent as the Kerr Collar, but they were capable of reducing loads in some configurations.

In a side impact, none of the collars substantially reduced loads in multiple configurations. Only the Kerr Collar reduced the lower neck moment. Again, this is due to the base of the helmet contacting the collar, restricting the range of motion. This movement restriction is most noticeable in the high speed video. The Cowboy Collar and Bullock Collar provided no side impact protection.

Through analyzing this data, it is obvious that the Kerr Collar performs differently than the other collars tested. The Kerr Collar is specifically designed to contact the base of the helmet, which restricts motion of the head and neck. The Cowboy Collar and Bullock Collar are designed to prevent hyperextension of the neck. Therefore, the Cowboy Collar and Bullock Collar only reduce loads in front impacts. Restriction of motion correlates with load reductions for each of the collars. In the future, manufacturers should consider restricting the motion of the head and neck in more orientations than just hyperextension when designing collars. This restriction of motion should lead toward distributing loads to the shoulders, rather than the head and neck.

It was also evident that the collars generally performed better when the shoulder pads were in the raised shoulder pad positions. This is mainly due to earlier contact with the neck collars.

6 CONCLUSION

A series of 48 tests were performed to assess the dynamic biomechanical effects of neck collars currently used in competitive football. Each neck collar was tested at two different impact speeds, at three different impact locations, and two different shoulder pad positions. With the top impact location, it was found that the Kerr Collar and Bullock Collar reduced head accelerations and force transmission through the neck, with the Kerr Collar producing greater reductions in force transmission. The Cowboy Collar produced no reductions in a top impact. With the front impact location, all the collars reduced lower neck moment, while the Kerr Collar was also capable of reducing the lower neck force and upper neck moment. With the side impact location, the Kerr Collar produced the greatest lower neck moment reductions. These reductions in loads correlate with how much each collar restricted the motion of the head and neck. Overall, the collars performed better when the shoulder pads were in the raised configuration.

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