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Designing a Mechanical Linkage Capable of Decreasing Force Transfer from the Facemask to the Protective Helmet when Loading Occurs

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DESIGNING A MECHANICAL LINKAGE CAPABLE OF DECREASING FORCE TRANSFER FROM THE FACEMASK TO THE PROTECTIVE HELMET WHEN LOADING OCCURS

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ABSTRACT

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Introduction

Sports that involve extensive personal contact have a high incidence of injury. The introduction of regulations mandating the use of personal protective equipment in these sports is the most common injury control strategy (Marshall et al., 2002). Negligible attention has been paid to the mechanical linkage between the facemask and helmet as a means of reducing force transfer from the facemask, through the helmet, and to the head and or neck of the athlete.

Methods

A novel prototype mechanical linkage of reasonable simplicity that provides 360° of freedom in motion capable of decreasing force transfer from the facemask to the protective helmet when loading occurs was designed. Force was applied at three angulations to the long axis of the a control and prototype mechanical linkage, under both compressive and tensile force, generating six experimental groups: Tension at 0°, Tension at 45°, Tension at 90°, Compression at 0°, Compression at 45°, and Compression at 90°. For each experimental group, the force transferred from the facemask connector to the helmet connector and deflection of the mechanical linkage at failure was evaluated.

Results

For each condition measured under both compressive and tensile force; maximum force transfer within the limits of the theoretical range of motion, force transfer at failure and linkage deflection at failure statistically significant differences between the control and prototype groups were observed with a *t* test for independent samples with unequal variance ($p < 0.001$), $\alpha = 0.05$.

Conclusion

When compared to currently available designs, the prototype mechanical linkage designed and tested as part of this project is of reasonable simplicity, displays increased flexibility and provides 360° of freedom in motion. Under compressive and tensile forces, force transfer from the facemask component to helmet component was decreased significantly.

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CHAPTER 1: INTRODUCTION

Background and Significance

The human head houses the sensory apparatus for hearing, vision, smell, taste and related lingual and labial sensations. In order to function optimally, these sensory organs must be able to scan the environment and be delivered towards objects of interest. The cervical spine supports this sensory platform, and moves and orientates it in three-dimensional space (Bogduk & Mercer, 2000).

Injury to the head and or neck can happen to an athlete at any level of participation, ranging from unsupervised activities to organized contact and collision sports. These injuries may occur in a vast array of sports, including but not limited to football (Vaccaro et al, 2002).

According to the National Spinal Cord Injury Statistical Center, approximately 12,000 new cases of spinal cord injuries occur each year, with sports-related events causing approximately 7.6% of the injuries (Zahir & Ludwig, 2010). Football is associated with the largest number of overall catastrophic cervical spine injuries according to the National Center for Catastrophic Sports Injury Research (Boden, Tacchetti, Cantu, Knowles, & Mueller, 2006). In relation, high-school and collegiate athletes endure an average of 7.23 direct catastrophic head injuries per year (Boden, Tacchetti, Cantu, Knowles, & Mueller, 2007) and nearly 85% of all football-related fatalities, between 1945 and 1994, resulted from head and cervical spine injuries (Zahir & Ludwig, 2010). The incidence of complete quadriplegia among high school and college football athletes has been reported to be as high as 2.5 per 100,000 (Vaccaro et. Al, 2002).

The inability of the nervous system to recover significant function following severe trauma (Torg, 1993), combined with the approximately 1.5 million high school and middle school athletes and more than 75,000 collegiate athletes participating in football each year (Zahir & Ludwig, 2010); generates an interest in the enhancement of player safety through advances in equipment technology.

A great deal of attention has been given to the protection afforded by helmets in football. Helmets decrease the potential for traumatic brain injury following a collision by reducing the acceleration of the head upon impact; by this means decreasing both the brain-skull collision, as well as the sudden deceleration induced axonal injury (Daneshvar et al, 2011). Extensive research and development with regard to energy absorbing material within helmets, which act by compressing to absorb force during a collision and slowly restoring to its original shape, thereby prolonging the duration of the collision while reducing the total momentum transferred to the head has been conducted (Daneshvar et al, 2011).

In contrast, negligible attention has been paid to the mechanical linkage between the facemask and helmet as a means of reducing force transfer from the facemask, through the helmet, and to the head and or neck of the athlete.

Purpose of Study

This study aims to explore whether it is possible to design a novel mechanical linkage of reasonable simplicity that provides 360° of freedom in motion with the objective of decreasing force transfer from the facemask to the protective helmet when loading occurs.

Research Questions and Hypotheses

Research Question 1

Is it possible to design a novel mechanical linkage of reasonable simplicity that provides 360° of freedom in motion capable of decreasing force transfer from the facemask to the protective helmet when loading occurs?

Null Hypothesis A (H_{0a}): Designing a novel mechanical linkage of reasonable simplicity that provides 360° of freedom in motion is not possible.

Alternate Hypothesis A (H_{1a}): Designing a novel mechanical linkage of reasonable simplicity that provides 360° of freedom in motion is possible.

Research Question 2

Can significant decreases in force transfer be obtained when compressive (frontal impact) forces are applied to the prototype mechanical linkage?

Null Hypothesis B (H_{0b}): The prototype mechanical linkage will not decrease measured force transfer from the facemask component to helmet component when compressive force is applied at 0^0, 45^0, and or 90^0. That is, for mean force transfer:

$$M_{C0} = M_{P0}$$

$$M_{C45} = M_{P45}$$

$$M_{C90} = M_{P90}$$

Alternate Hypothesis (H_{1b}): The prototype mechanical linkage will decrease measured force transfer from the facemask component to helmet component when compressive force is applied at 0^0, 45^0, and 90^0. That is, for mean force transfer:

$$M_{C0} \neq M_{P0}$$

$$M_{C45} \neq M_{P45}$$

$$M_{C90} \neq M_{P90}$$

Research Question 3

Can significant decreases in force transfer be obtained when tensile (pulling) forces are applied to the prototype mechanical linkage?

Null Hypothesis C (H_{0c}): The prototype mechanical linkage will not decrease measured force transfer from the facemask component to helmet component when tensile force is applied at 0^0, 45^0, and 90^0. That is, for mean force transfer:

$$M_{C0} = M_{P0}$$

$$M_{C45} = M_{P45}$$

$$M_{C90} = M_{P90}$$

Alternate Hypothesis C (H_{1c}): The prototype mechanical linkage will decrease measured force transfer from the facemask component to helmet component when tensile forces are applied at 0^0, 45^0, and 90^0. That is, for mean force transfer:

$$M_{C0} \neq M_{P0}$$

$$M_{C45} \neq M_{P45}$$

$$M_{C90} \neq M_{P90}$$

CHAPTER 2: LITERATURE REVIEW

Protective Sports Equipment

Personal protective sports equipment acts to buffer the major body segments; such as the face, head, neck, arms, legs, chest, shoulders, abdomen and legs from injurious assault during physical contact.

Significance

Sports that involve extensive personal contact have a high incidence of injury. The introduction of regulations mandating the use of personal protective equipment in these sports is the most common injury control strategy (Marshall et al., 2002). An international epidemiological study conducted by Marshall et al. in 2002 found that sports mandating the use of personal protective equipment had an injury rate approximately one-third the rate of sports that do not mandate personal protective equipment. Furthermore, a pattern of decreasing risk with increasing level of protective equipment across body site was observed. The most noteworthy effect was related to head injuries, in which sports requiring personal protective equipment showed an injury rate one-tenth of those that did not (Marshall et al., 2002).

The United States Centers for Disease Control and Prevention asserts that participation in organized sports is on the rise, with approximately 30 million children and adolescents participating in youth sports in the United States alone (Weisenberger, 2014). Accordingly, an emphasis on the utilization of proper personal protective equipment in sports equipment has assumed a prominent role.

In the discipline of Orthodontics and Dentofacial Orthopedics, protection of the head, neck and face is of notable importance. Each year, in April, the American

Association of Orthodontists promotes National Facial Protection Month, aimed at reminding athletes that wearing appropriate personal protective equipment at every practice and game during recreational and organized sports will help them remain safe. In many contact sports; including football, hockey, baseball, softball, lacrosse and others, the use of facemasks, fastened to a helmet are utilized to help accomplish this goal.

Rules and Regulations

The use of facemasks, fastened to protective helmets of various designs, is now mandated by most professional leagues in which extensive personal contact occurs during gameplay. Often, all youth or amateur subsidiaries of these professional leagues implement the same or similar rules. The following professional leagues have mandated the use of facemasks by some or all participants:

National Football League. Requires that "players must wear the equipment and uniform apparel listed below,…helmet…[with] facemask attached. Facemasks must not be more than 5⁄8-inch in diameter and must be made of rounded material..." (Official NFL Rules, 2013).

Major League Baseball. Requires that "all catcher's wear a catcher's protective helmet, while fielding their position" (Official MLB Rulebook, 2012). According to the National Operating Committee on Standards for Athletic Equipment "all…[catcher's] helmets must be…with the faceguard (mask) attached and shall be mounted on a catcher's helmet according to the manufacturer's instructions" (NOCSAE Baseball Helmets, 2012).

National Hockey League. Requires that "protective masks of a design approved by the League must be worn by goalkeepers" (Official NHL Rules, 2012).

United States Lacrosse. Requires the use of "…mouth guards, arm pads, gloves, shoulder pads, and NOCSAE Helmets" (Official Lacrosse Rules, 2014). The National Operating Committee on Standards for Athletic Equipment in turn, states that "all…[lacrosse] helmets must be…with a compatible faceguard (mask) that has been certified to meet the NOCSAE standard..." (NOCSAE Lacrosse Helmets, 2012).

Facemask History

The introduction date of facemasks as a component of the sports protective equipment repertoire differs based on the allegiance of the sports historian consulted. Popular football lore contends that the helmet manufacturer Riddell created the first modern face mask for Otto Graham, a quarterback with the Cleveland Browns, in 1953 (Bird, 2011). Baseball historians attribute the idea to Fred Thayer of the Harvard University Baseball Club, in 1875, and some say the catcher's mask might have been first worn by Jim Tyng, in 1876, when he modified a fencing mask (Epic Sports, 2014). Hockey aficionados believe that the first facemask was worn by Queen's University goaltender Elizabeth Graham to protect her teeth (USA Hockey, 1999).

Nevertheless, it has been definitively established that improvised facemasks were used as early as the 1920s. In the early years, players often wore nose-guards constructed from leather as their only means of facial protection (Bird, 2011), and there even exists an old helmet with a barbed wire facemask (Worrell, 2014). By the 1930s, facemasks had evolved to cover the entire face with holes cut out for the eyes and mouth.

Since they were made widely available in the 1950s, many manufacturers have produced facemasks, including but not limited to: Adams, Dungard, MacGregor, Marietta, Riddell, Rawlings, Schutt, and Wilson (Worrell, 2014). Countless facemask designs have been explored and employed over time; however, the mechanical linkage responsible for fastening the facemask to the helmet has remained largely unchanged. Historically, facemasks were rigidly fixed to the helmet directly via standard screws, indirectly via loop straps in combination with standard screws and less commonly directly via leather straps (Worrell). Currently, the most common method of attachment remains the loop strap, attached via standard screw, as evaluated in the coming text.

Current Research

As stated previously, a great deal of attention has been given to the protection afforded by helmets in football. Helmets decrease the potential for traumatic brain injury following a collision by reducing the acceleration of the head upon impact; by this means decreasing both the brain-skull collision, as well as the sudden deceleration induced axonal injury (Daneshvar et al, 2011). Extensive research and development with regard to energy absorbing material within helmets, which act by compressing to absorb force during a collision and slowly restoring to its original shape, thereby prolonging the duration of the collision while reducing the total momentum transferred to the head has been conducted (Daneshvar et al, 2011).

In contrast, negligible attention has been paid to the mechanical linkage between the facemask and helmet as a means of reducing force transfer from the facemask, through the helmet, and to the head and or neck of the athlete.

CHAPTER 3: MATERIALS AND METHODS

Novel Mechanical Linkage Design

Extensive research, development, trial and error with the intent to design a novel mechanical linkage of reasonable simplicity that provides 360° of freedom in motion capable of decreasing force transfer from the facemask to the protective helmet when loading occurs generated a prototype for the mechanical linkage with three basic components: 1) Helmet Connector 2) Facemask Connector, and 3) Two-way Elastomeric Receptacle; as seen in Figure 1.

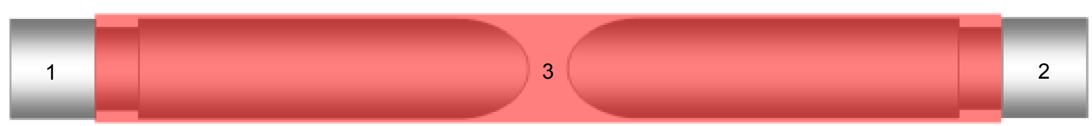

Figure 1. Mechanical Linkage Diagram: 1) Helmet Connector, 2) Facemask Connector, 3) Elastomeric Sleeve

It was not the goal of this project to determine the ideal materials to act as said components; but instead to establish a design concept that meets the aforementioned criterion using basic ubiquitous materials. In addition, the design was to be of such a nature that component materials could be interchanged to improve the performance of the mechanical linkage with relative ease, while remaining in compliance with the structural and material standards set forth for facemasks by regulatory agencies.

Helmet and Facemask Connector

As a point of reference, the helmet and facemask connector was designed to comply with the structural and material standards set forth for facemasks by National Football League. According to the official rulebook of the National Football League and Commissioner Roger Goodell, facemasks must not be more than 5/8-inch in diameter and must be made of rounded material; and transparent materials are prohibited (2013). The facemask connector and helmet connector are structurally identical in all dimensions, each fastened to the facemask and helmet respectively. The connectors were fabricated from stainless steel, due to its acceptable physical properties and low coefficient of

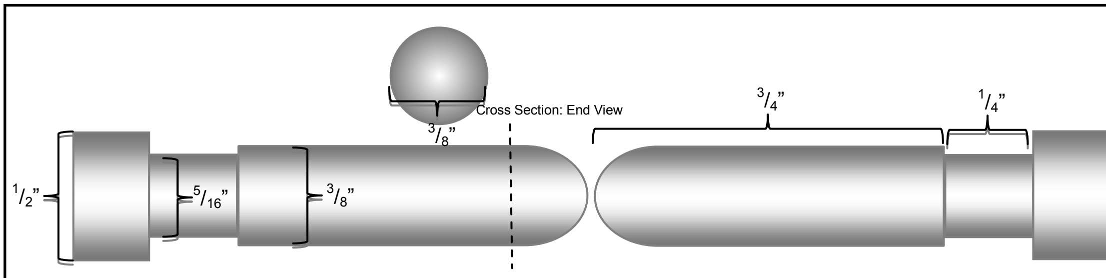

Figure 2. Helmet and Facemask Connector Dimensions

frictional resistance (Proffit, 2004). As seen in Figure 2, at the point of approximation, the connectors are half-spherical in shape, naturally tapering into the shape of a cylinder of diameter 0.375 inches to a length of 0.75 inches. At this point, a 90^0 ledge is created by decreasing the diameter of the cylinder to 0.3125 inches for to an additional length of 0.25 inches; the ledge functions as a retention barb for the elastomeric receptacle. A second 90^0 ledge is created by increasing the diameter of the cylinder to 0.5 inches for an unspecified distance; the ledge serves as a buttress for the end of the elastomeric receptacle, as can be delineated in Figure 1 above and Figure 3 below. The portion distal to the second 90^0 ledge of the connector serves as an area for fastening to the helmet for facemask respectively.

Elastomeric Receptacle

The elastomeric receptacles were fabricated from standard rubber latex surgical tubing, due to its acceptable physical properties. Surgical tubing has the shape of a hollow cylinder. For this application, tubing of the following dimension were used: inside

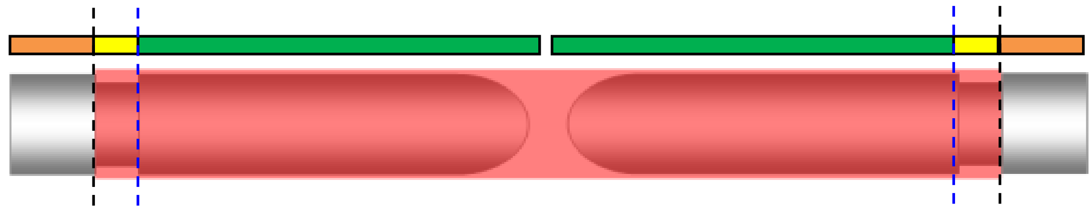

Figure 3. Interface Diagram:

Green Sector: connector O.D. 0.375 inches approximating elastomeric receptacle I.D. of 0.25 inches

Blue Dashed-Line: 90^0 ledge, diameter of the cylinder decrease from 0.375 to 0.3125 inches; retention barb

Yellow Sector: connector O.D. 0.3125 inches approximating elastomeric receptacle I.D. of 0.25 inches

Black Dashed-Line: 90^0 ledge, diameter of the cylinder increases from 0.3125 to 0.5 inches; flush buttress

Orange Sector: area for fastening connectors to helmet and facemask respectively

diameter (I.D.) or lumen size of 0.25 inches, outside diameter (O.D.) of 0.5 inches, leaving a wall thickness of 0.125 inches. The total length of the elastomeric receptacle was 2.0 inches, allowing buttressing of the elastomeric tubing to the distal 90^0 ledge, creating a flush junction, as seen in Figure 3.

Connector, Receptacle Interface

Interface relationships of the varying inside and outside diameters of the elastomeric receptacle and connector are illustrated in Figure 3. With the approximation of the connectors as an origin, areas of note are the proximal segments in which the connector O.D. is 0.375 inches and the elastomeric receptacle I.D. is 0.25 inches, creating a friction grip interface. Next, at the point of the proximal 90^0 ledge, 0.75 inches from the approximation of the connectors, the diameter of the connector cylinder decreases from to 0.3125 inches, effectively creating a retention barb for the elastomeric receptacle. Lastly, at the distal 90^0 ledge the outside diameter of the connector cylinder and elastomeric receptacle are equal, creating a flush buttress for the end of the elastomeric receptacle.

Control Data

Information regarding material composition and physical properties of traditional

Figure 4. Control Specimen

rigid connectors is not readily available from the manufacturer or in the literature. Force transfer during loading, from the facemask connector to helmet connector, represents the theoretical force that could be transferred to the head and neck of an athlete under impact conditions during an athletic event.

Control data representing force transfer during loading was obtained by applying compressive and tensile force to traditional rigid receptacles; Schutt Armorguard Elite Facemask Loop Strap Clips (Item #: 15002221), Figure 4. Force was applied at three angulations to the long axis of the control mechanical linkage, under both compressive and tensile force, generating six experimental control groups: Control Force Transfer in Tension at 0°, Control Force Transfer in Tension at 45°, Control Force Transfer in Tension at 90°, Control Force Transfer in Compression at 0°, Control Force Transfer in Compression at 45°, and Control Force Transfer in Compression at 90°; represented diagrammatically in Figure 5.A below. The maximum force value endured by the helmet connector via the traditional rigid receptacle, Schutt Armorguard Elite Facemask Loop Strap Clips (Item #: 15002221) at the full theoretical range of motion of the prototype mechanical linkage and at failure established the value for potential force that may be

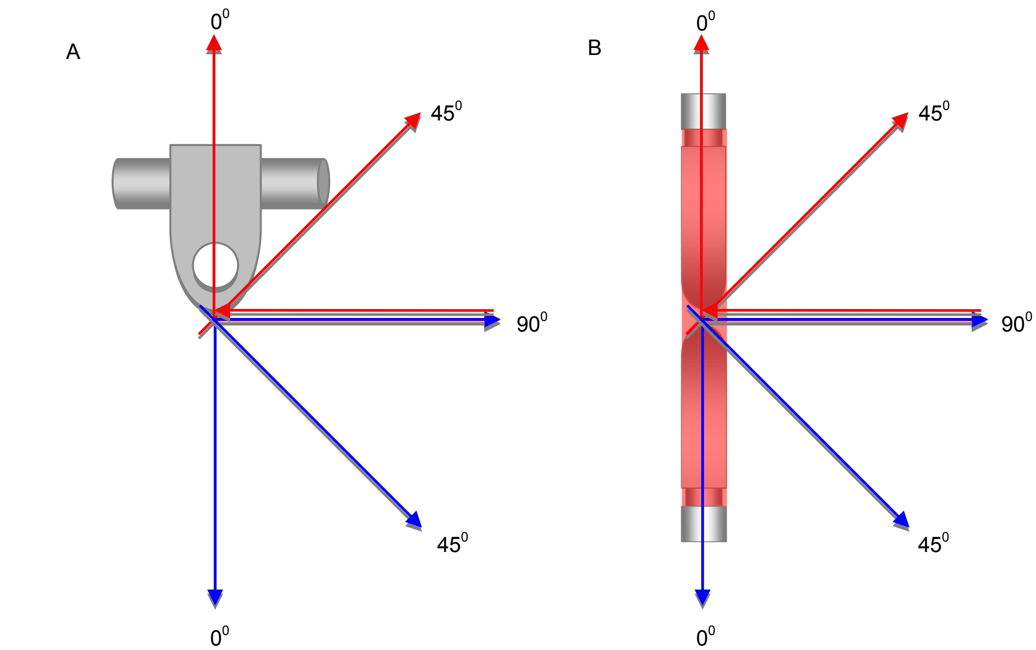

Figure 5. A) Control and B) Prototype Group Diagram

Red: Compression

Blue: Tension

transferred from the facemask connector to the helmet connector at each angulation. Each control group was tested 5 times (n = 5) to establish control statistics.

Prototype Data

Prototype data representing force transfer during loading was obtained by applying compressive and tensile force to the novel mechanical linkage design receptacles; outlined above. Force was applied at three angulations to the long axis of the prototype mechanical linkage, under both compressive and tensile force, generating six experimental prototype groups: Prototype Force Transfer in Tension at 0°, Prototype Force Transfer in Tension at 45°, Prototype Force Transfer in Tension at 90°, Prototype

Force Transfer in Compression at 0°, Prototype Force Transfer in Compression at 45°, and Prototype Force Transfer in Compression at 90°; represented diagrammatically in Figure 5.B above. The maximum force value endured by the helmet connector, via the novel mechanical linkage design receptacle, at the full theoretical range of motion of the prototype mechanical linkage and at failure established the value for potential force that may be transferred from the facemask connector to the helmet connector at each angulation. Each prototype group was tested 5 times (n = 5) to establish prototype statistics. The variation in maximum force experienced by the football helmet connector via the prototype receptacle, as reference to the control statistics, represents the potential change in force that could be transferred to the head and neck of an athlete under impact conditions during an athletic event.

Specimen Testing and Data Collection Procedure

The instrumentation used for monitoring experimental cycles of compressive and tensile force transfer was a Tinius Olsen S Series Materials Testing Machine, with

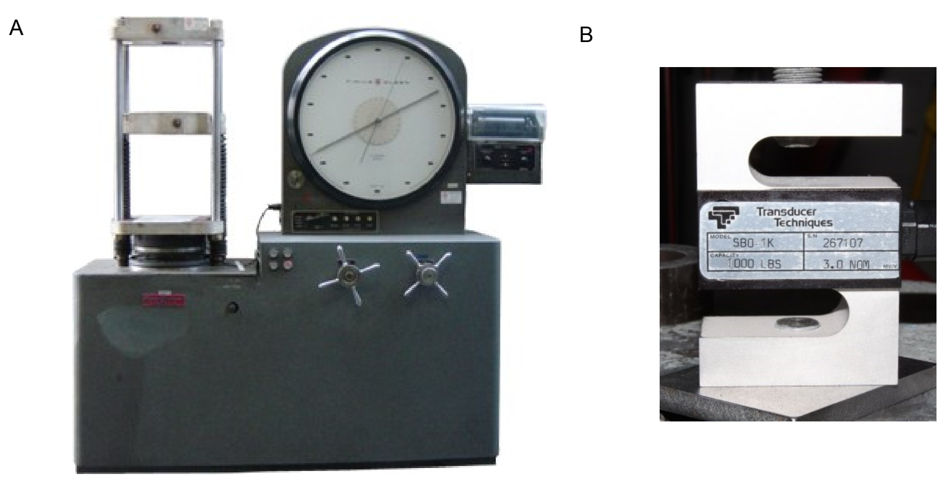

Figure 6. Experimental Instrumentation: A) Tinius Olsen Machine (Stock Photo) and B) Load Cell

adapted 1000lb load cell (Transducer Techniques, Model SB0-1K, 267107), as seen in Figure 6. Tensile and compressive force was recorded simultaneously with displacement, which was monitored by an adapted Extensometer (Epsilon, Model 3540-200T-ST, Serial Number E87707), shown in Figure 6.

Control and prototype receptacle specimens were fastened into custom-fabricated jigs simulating the helmet connector and facemask connectors, at the aforementioned angulations for both the control and prototype groups. The custom-fabricated jigs were secured with the appropriate hardware to the base of the Tinius Olsen S Series Materials Testing Machine, with adapted 1000lb load cell (Transducer Techniques, Model SB0-1K, 267107) representing the helmet connector and action arm of the Tinius Olsen S Series Materials Testing Machine, representing the facemask connector. The active arm of the Tinius Olsen S Series Materials Testing Machine, representing the facemask connector, was advanced at a rate of 0.05 inches per minute for all test groups.

Monitored data was interpreted and logged from the Tinius Olsen S Series Materials Testing Machine, with adapted 1000lb load cell (Transducer Techniques, Model SB0-1K, 267107) and Extensometer (Epsilon, Model 3540-200T-ST, Serial Number E87707) via a P3 Strain Indicator and Recorder in conjunction with associated software, creating simple text files for each specimen that was later transcribed into Microsoft Excel for data manipulation and analysis.

Statistical Analysis

This study used a normal materials sampling design to evaluate the force transfer through a mechanical linkage in compressive and tensile loading. A preliminary test of variances was not performed, because literature supports the assertion that an unequal variances t test performed without an initial comparison of variances has high power in situations in which it is not known whether the underlying population variances are equal, rendering the initial check ineffective and or unnecessary (Pagano & Gauvreau, 1993). It was assumed that both control and prototype samples were drawn from Gaussian populations, but not assumed that the populations had equal standard deviations.

As such, to compare the independent control and prototype samples, data was analyzed with an unequal variance t test, also known as the Welch t test, at a significance level of 0.05 ($\alpha = 0.05$) for six experimental groups, three conditions: Compression at 0°, Compression at 45°, Compression at 90°, Tension at 0°, Tension at 45° and Tension at 90°. For each of these six experimental groups, the following three conditions were evaluated statistically: maximum force transfer within the limits of the theoretical range of motion, maximum force at failure and deflection of the mechanical linkage at failure. In addition, to enumerate the accuracy of the mean of each experimental group, confidence intervals were constructed. All data were analyzed for statistically differences using Microsoft Excel Analysis Toolpak Add-On.

CHAPTER 4: RESULTS

Compression at 0°

Force transfer testing in compression at 0 degrees to the long axis of the mechanical linkage for control and prototype specimen (N = 10) was carried out with custom-fabricated fixtures. Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under zero degree compressive stress is located in Table 1.

Table 1 Compression at 0° (Degrees)

	Control			Prototype		
	ROM Force	Fail Force	Fail Distance	ROM Force	Fail Force	Fail Distance
Specimen 1	191	235	1.367	16	23	3.469
Specimen 2	209	255	1.264	17	25	3.656
Specimen 3	192	205	1.459	16	22	3.499
Specimen 4	187	219	1.332	19	26	3.438
Specimen 5	201	292	1.575	18	24	3.938

Note: ROM: theoretical range of motion of prototype joint; Fail: Failure

Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under compressive stress at zero degrees to the long axis of the linkage. Force data reported in pounds-force (lbf) and deflection data reported in inches (in).

As it pertains to maximum force transfer within the limits of the theoretical range of motion for control specimen; a mean of 196 pounds-force (SD = 8.89), with a 95% confidence interval of [188, 204] was observed (n = 5). For prototype specimen; a mean of 17.2 pounds-force (SD = 1.30), with a 95% confidence interval of [16.1, 18.3] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 44.51$. Significant differences in data for the control (M = 196, SD = 8.89) and prototype (M = 17, SD = 1.30) specimen were observed: $t(4) = 44.51$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, see Figure 7.

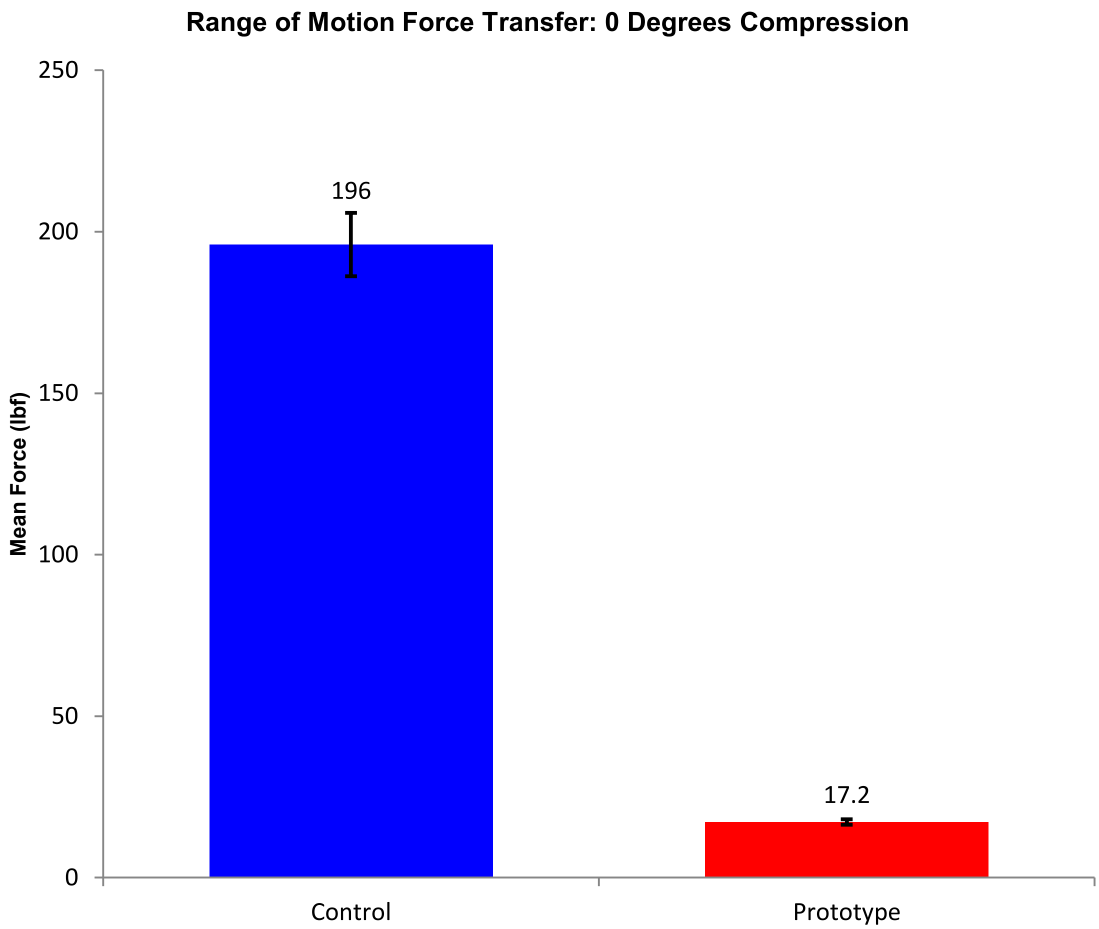

Figure 7: Significant differences in data for the control (*M* = 196, *SD* = 8.89) and prototype (*M* = 17, *SD* = 1.30) specimen were observed: $t(4) = 44.51$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to maximum force transfer at failure for control specimen; a mean of 241 pounds-force (*SD* = 33.9), with a 95% confidence interval of [211, 271] was observed ($n = 5$). For prototype specimen; a mean of 24 pounds-force (*SD* = 1.58), with a 95% confidence interval of [22.6, 25.4] was observed ($n = 5$). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 14.29$. Significant differences in data for the control (*M* = 241, *SD* = 33.9) and prototype (*M* = 24, *SD* = 1.58) specimen were observed: $t(4) = 14.29$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, Figure 8.

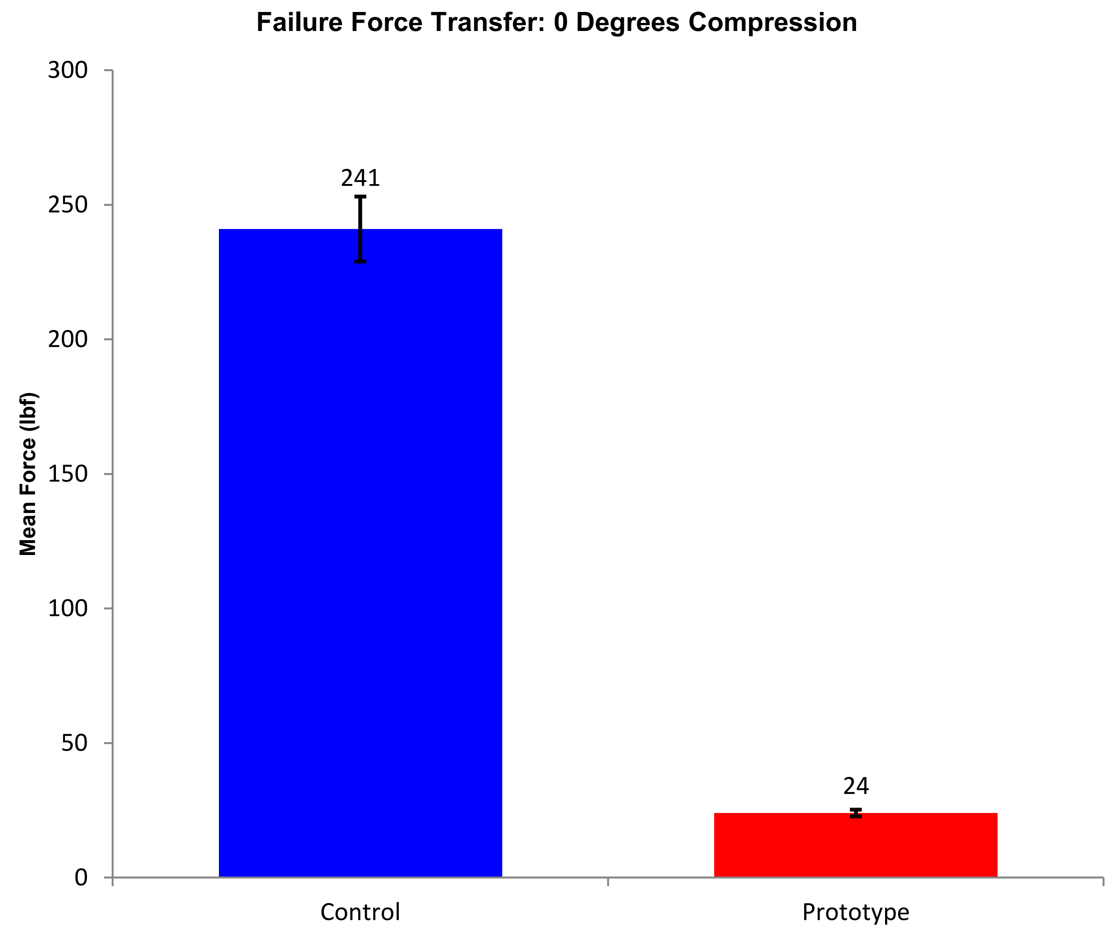

Figure 8: Significant differences in data for the control (*M* = 241, *SD* = 33.9) and prototype (*M* = 24, *SD* = 1.58) specimen were observed: $t(4) = 14.29$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to deflection at failure for control specimen; a mean of 1.39 inches (*SD* = 0.12), with a 95% confidence interval of [1.29, 1.51] was observed (*n* = 5). For prototype specimen; a mean of 3.60 inches (*SD* = 0.21), with a 95% confidence interval of [3.42, 3.78] was observed (*n* = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 20.55$. Significant differences in data for the control (*M* = 1.39, *SD* = 0.12) and prototype (*M* = 3.60, *SD* = 0.21) specimen were observed: $t(4) = 20.55$, $p < 0.001$; one-tailed t distribution critical value = 1.94, two-tailed t distribution critical value = 2.45, Figure 9.

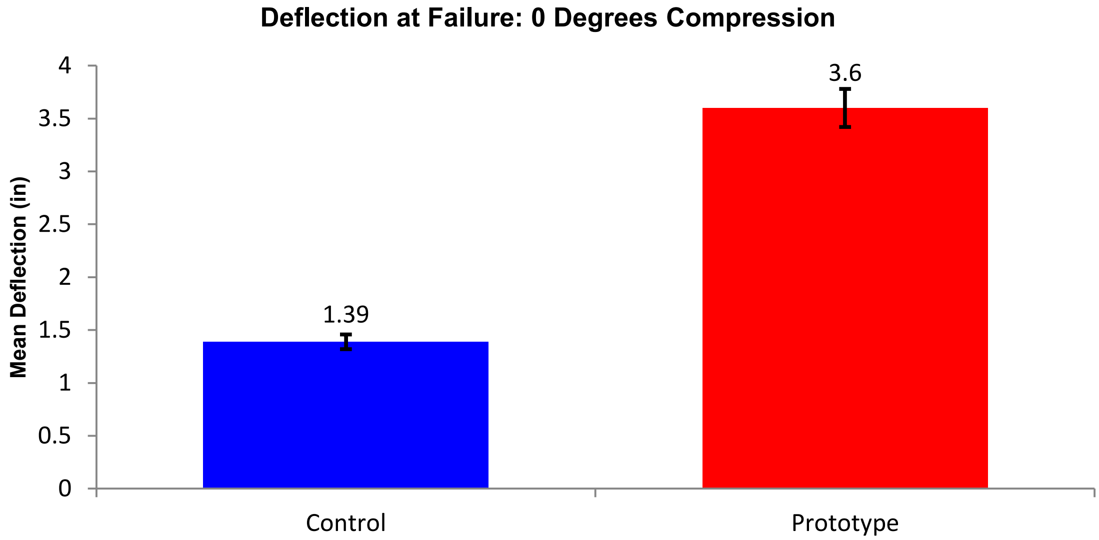

Figure 9. Significant differences in data for the control (M = 1.39, SD = 0.12) and prototype (M = 3.60, SD = 0.21) specimen were observed: $t(4) = -20.55$, $p < 0.001$; one-tailed t distribution critical value = 1.94, two-tailed critical value = 2.45.

Compression at 45°

Force transfer testing in compression at 45 degrees to the long axis of the mechanical linkage for control and prototype specimen (N = 10) was carried out with custom-fabricated fixtures. Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under forty-five degree compressive stress is located in Table 2.

Table 2: Compression at 45^{o} (Degrees)

	Control			Prototype		
	ROM Force	Fail Force	Fail Distance	ROM Force	Fail Force	Fail Distance
Specimen 1	214	214	0.709	13	23	3.469
Specimen 2	211	211	0.831	13	25	3.656
Specimen 3	205	205	0.831	12	22	3.499
Specimen 4	192	192	0.881	13	26	3.438
Specimen 5	182	182	0.983	13	24	3.938

Note: ROM: theoretical range of motion of prototype joint; Fail: Failure

Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under compressive stress at forty-five degrees to the long axis of the linkage. Force data reported in pounds-force (lbf) and deflection data reported in inches (in).

As it pertains to maximum force transfer within the limits of the theoretical range of motion for control specimen; a mean of 201 pounds-force (*SD* = 13.5), with a 95% confidence interval of [189, 213] was observed (*n* = 5). For prototype specimen; a mean of 12.8 pounds-force (*SD* = 0.45), with a 95% confidence interval of [12.4, 13.2] was observed (*n* = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 31.17$. Significant differences in data for the control ($M = 201$, $SD = 13.5$) and prototype ($M = 12.8$, $SD = 0.45$) specimen were observed: $t(4) = 31.17$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, see Figure 10.

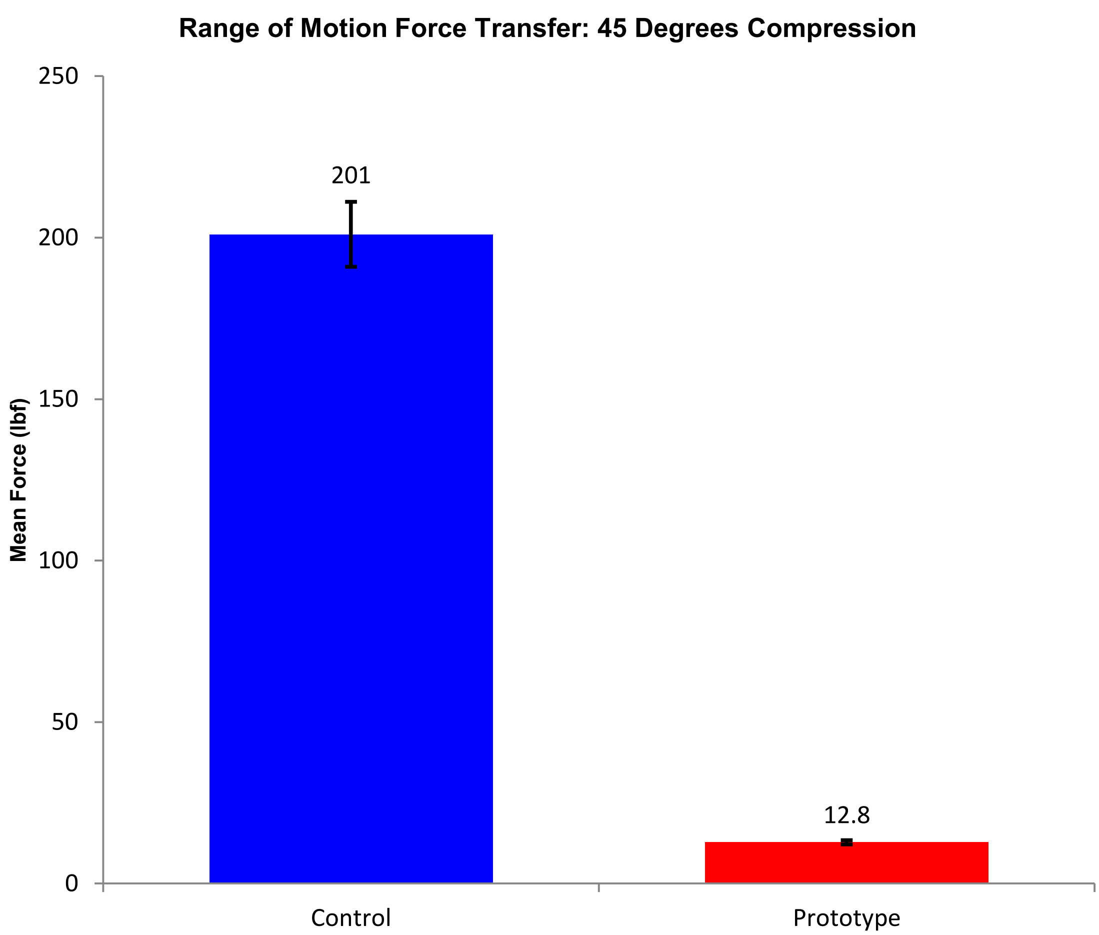

Figure 10: Significant differences in data for the control ($M = 201$, $SD = 13.5$) and prototype ($M = 12.8$, $SD = 0.45$) specimen were observed: $t(4) = 31.17$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to maximum force transfer at failure for control specimen; a mean of 201 pounds-force (*SD* = 13.5), with a 95% confidence interval of [189, 213] was observed ($n = 5$). For prototype specimen; a mean of 24 pounds-force (*SD* = 1.58), with a 95% confidence interval of [22.6, 25.4] was observed ($n = 5$). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 29.13$. Significant differences in data for the control ($M = 201$, $SD = 13.5$) and prototype (M = 24, SD = 1.58) specimen were observed: $t(4) = 29.13$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, Figure 11.

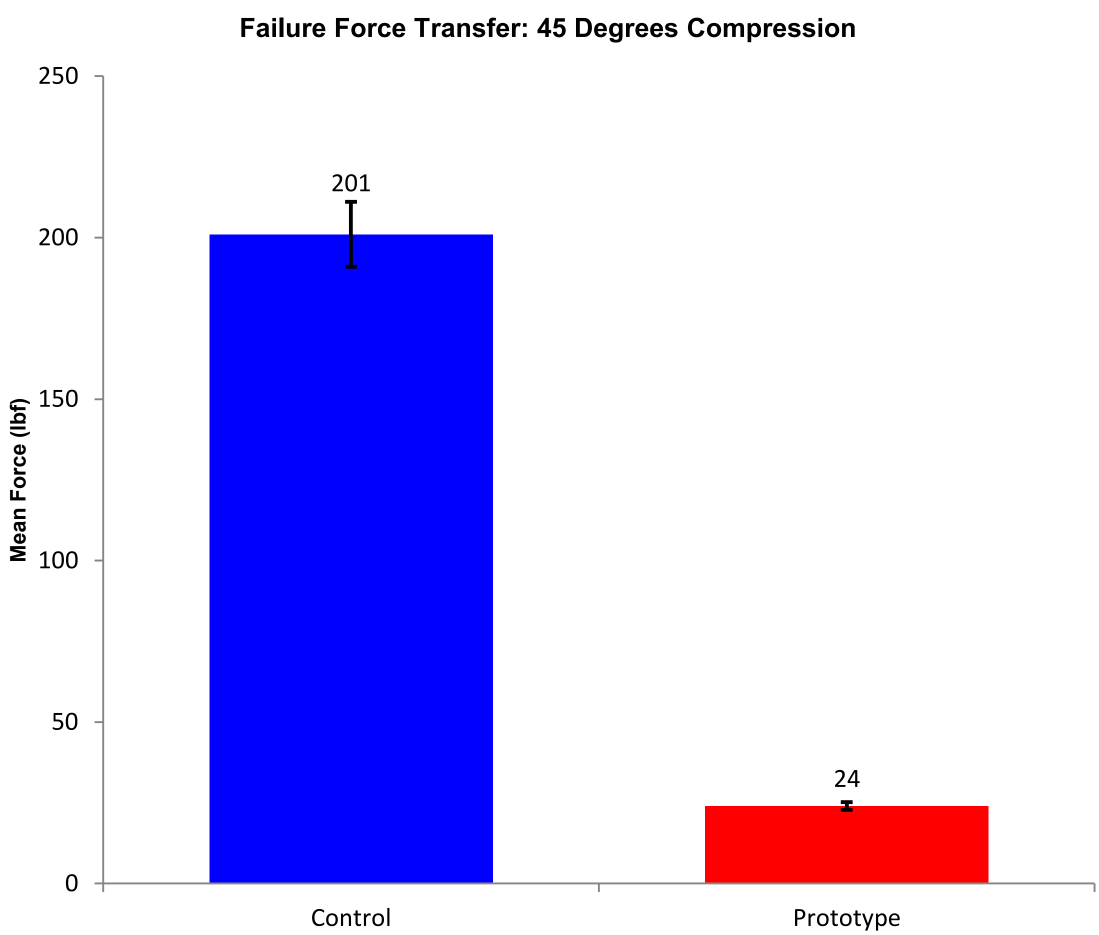

Figure 11. Significant differences in data for the control ($M = 201$, $SD = 13.5$) and prototype ($M = 24$, $SD = 1.58$) specimen were observed: $t(4) = 29.13$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to deflection at failure for control specimen; a mean of 0.85 inches (*SD* = 0.09), with a 95% confidence interval of [0.76, 0.93] was observed (n = 5). For prototype specimen; a mean of 3.60 inches (*SD* = 0.21), with a 95% confidence interval of [3.42, 3.78] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 26.85$. Significant differences in data for the control ($M = 0.85$, $SD = 0.09$) and prototype ($M = 3.60$, $SD = 0.21$) specimen were observed: $t(4) = 26.85$, $p < 0.001$; one-tailed t distribution critical value = 1.94, two-tailed t distribution critical value = 2.45, Figure 12.

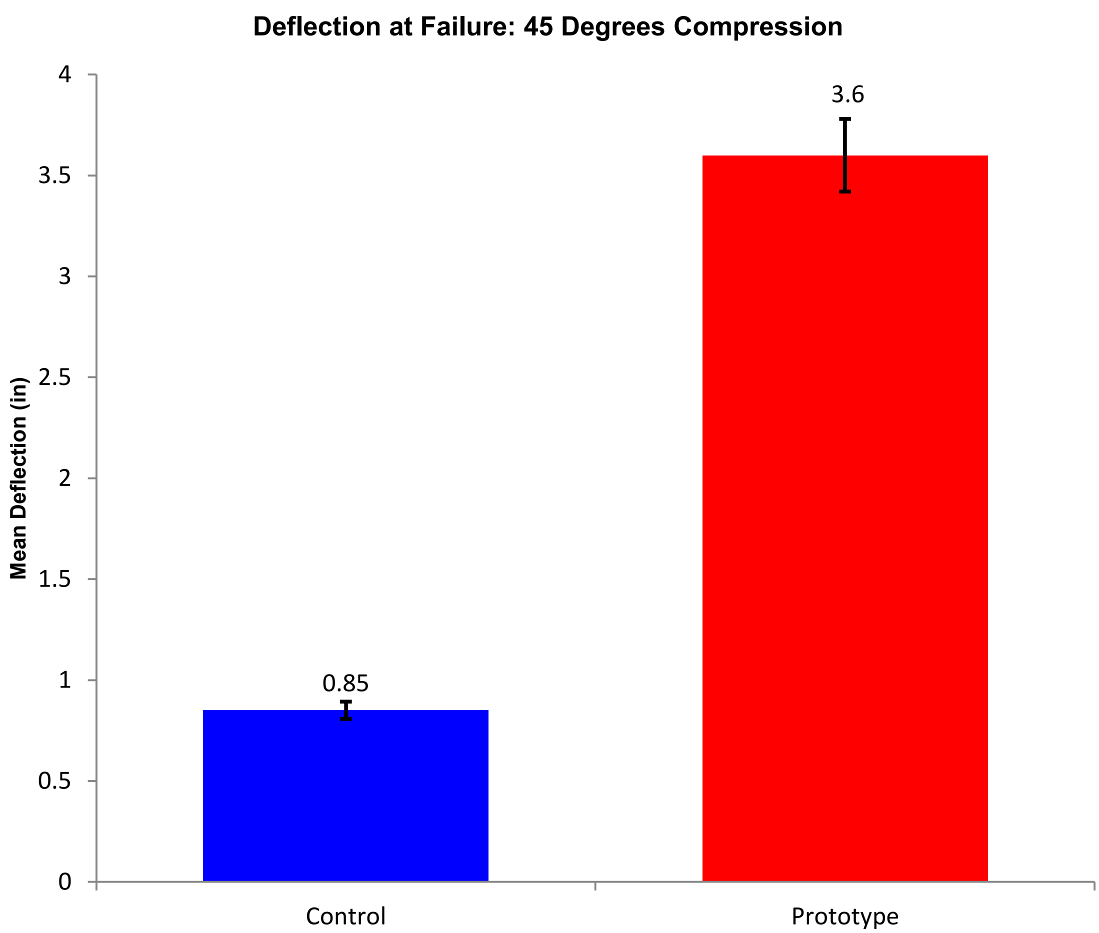

Figure 12. Significant differences in data for the control ($M = 0.85$, $SD = 0.09$) and prototype ($M = 3.60$, $SD = 0.21$) specimen were observed: $t(4) = -26.85$, $p < 0.001$; one-tailed t distribution critical value = 1.94, two-tailed critical value = 2.78.

Compression at 90°

Force transfer testing in compression at 90 degrees to the long axis of the mechanical linkage for control and prototype specimen (*N* = 10) was carried out with custom-fabricated fixtures. Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under ninety degree compressive stress is located in Table 3.

Table 3: Compression at 90^{o} (Degrees)

	Control			Prototype		
	ROM Force	Fail Force	Fail Distance	ROM Force	Fail Force	Fail Distance
Specimen 1	234	254	1.227	13	23	3.469
Specimen 2	256	263	1.134	13	25	3.656
Specimen 3	249	252	1.096	14	22	3.499
Specimen 4	243	279	1.253	13	26	3.438
Specimen 5	247	265	1.192	14	24	3.938

Note: ROM: theoretical range of motion of prototype joint; Fail: Failure

Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under compressive stress at ninety degrees to the long axis of the linkage. Force data reported in pounds-force (lbf) and deflection data reported in inches (in).

As it pertains to maximum force transfer within the limits of the theoretical range of motion for control specimen; a mean of 246 pounds-force (SD = 8.11), with a 95% confidence interval of [239, 253] was observed (n = 5). For prototype specimen; a mean of 13.4 pounds-force (SD = 0.55), with a 95% confidence interval of [12.9, 13.9] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 63.97$. Significant differences in data for the control (M = 246, SD = 8.11) and prototype (M = 13.4, SD = 0.55) specimen were observed: $t(4) = 63.97$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, see Figure 13.

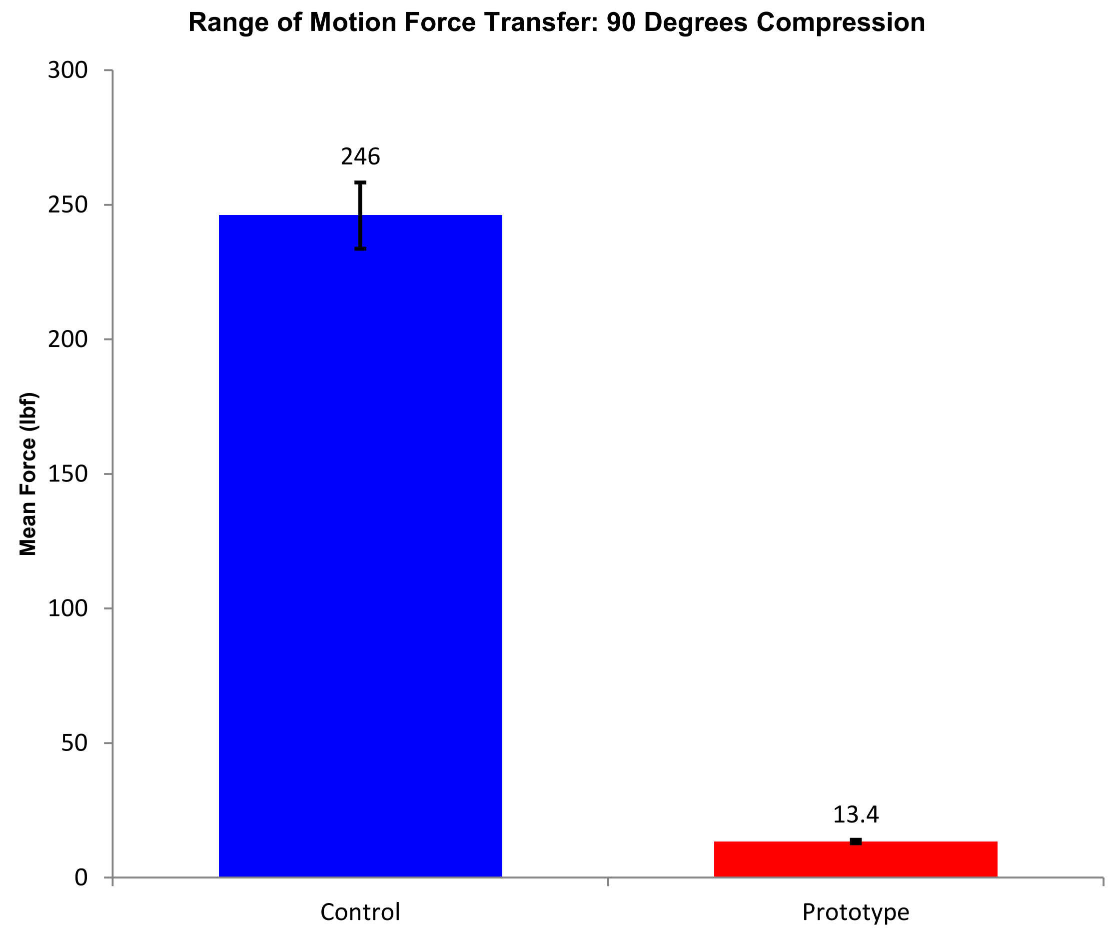

Figure 13: Significant differences in data for the control (M = 246, SD = 8.11) and prototype (M = 13.4, SD = 0.55) specimen were observed: $t(4) = 63.97$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to maximum force transfer at failure for control specimen; a mean of 263 pounds-force (SD = 10.7), with a 95% confidence interval of [253, 272] was observed (n = 5). For prototype specimen; a mean of 24 pounds-force (SD = 1.58), with a 95% confidence interval of [22.6, 25.4] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 49.16$. Significant differences in data for the control (M = 263, SD = 10.7) and prototype (M = 24, SD = 1.58) specimen were observed: $t(4) = 49.16$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, Figure 14.

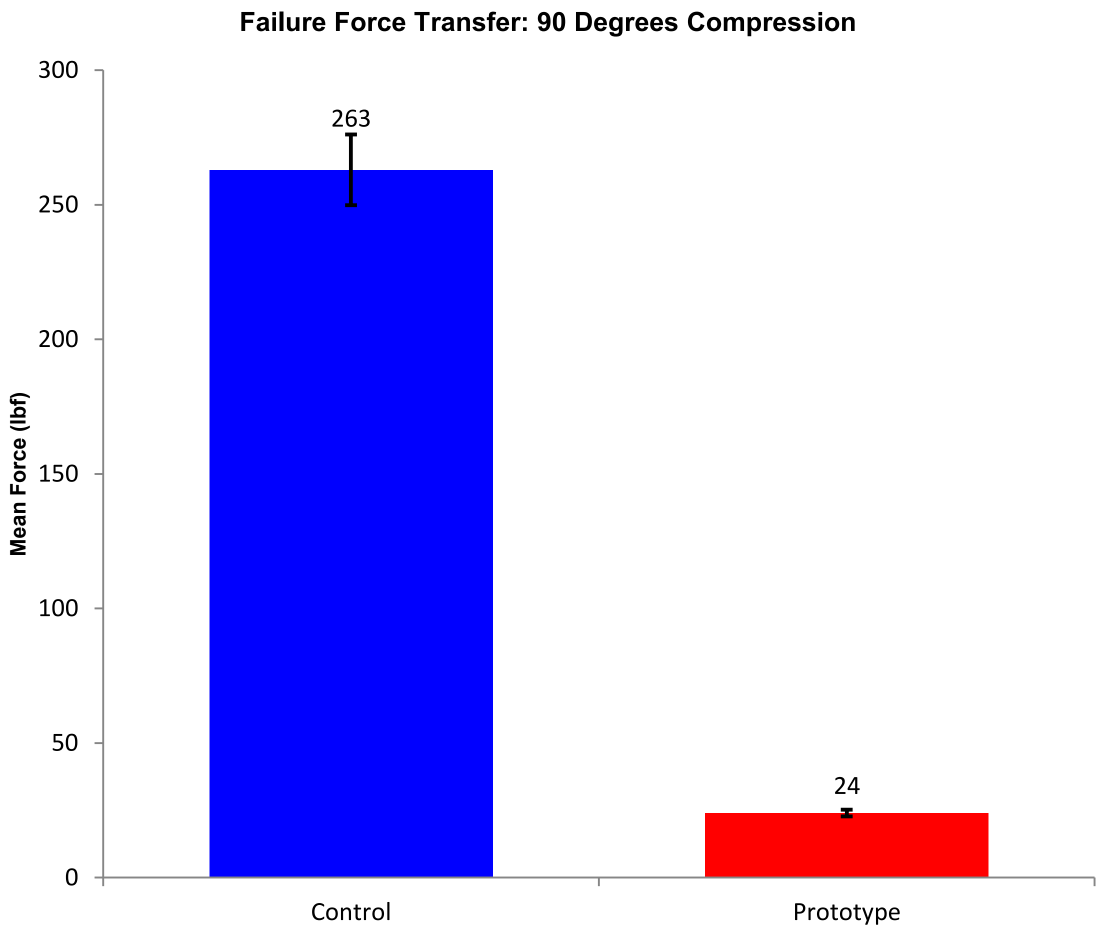

Figure 14. Significant differences in data for the control (M = 263, SD = 10.7) and prototype (M = 24, SD = 1.58) specimen were observed: $t(4) = 49.16$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to deflection at failure for control specimen; a mean of 1.18 inches (SD = 0.06), with a 95% confidence interval of [1.12, 1.24] was observed (n = 5). For prototype specimen; a mean of 3.60 inches (SD = 0.21), with a 95% confidence interval of [3.42, 3.78] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 24.97$. Significant differences in data for the control (M = 1.18, SD = 0.06) and prototype (M = 3.60, SD = 0.21) specimen were observed: $t(4) = 24.97$, $p < 0.001$; one-tailed t distribution critical value = 2.02, two-tailed t distribution critical value = 2.57, Figure 15.

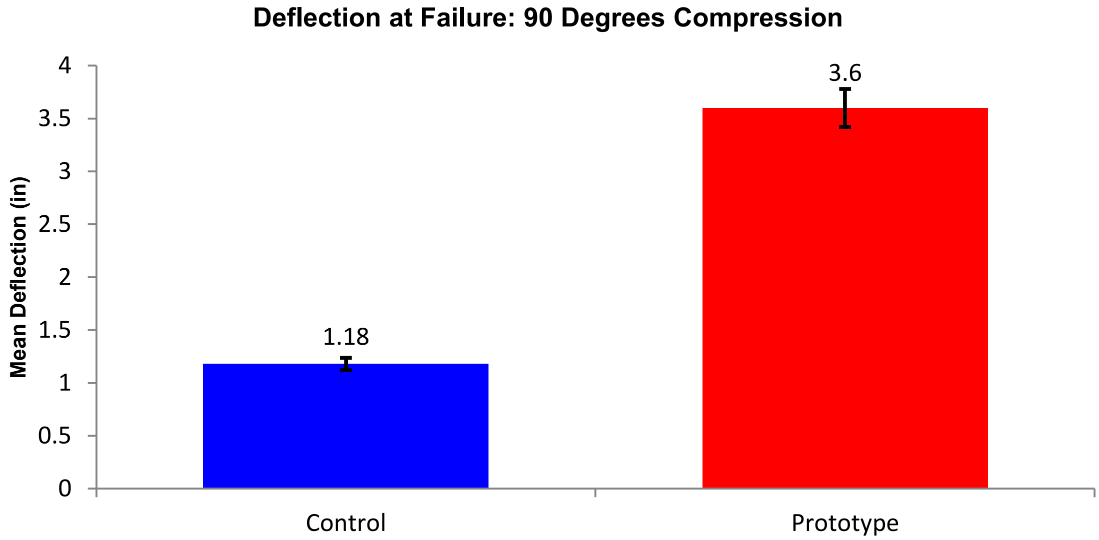

Figure 15. Significant differences in data for the control (*M* = 1.18, *SD* = 0.06) and prototype (*M* = 3.60, *SD* = 0.21) specimen were observed: $t(4) = -24.97$, $p < 0.001$; one-tailed t distribution critical value = 2.02, two-tailed critical value = 2.57.

Tension at 0°

Force transfer testing in tension at 0 degrees to the long axis of the mechanical linkage for control and prototype specimen (N = 10) was carried out with custom-fabricated fixtures. Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under zero degree tensile stress is located in Table 4.

Table 4: Tension at 0° (Degrees)

	Control			Prototype		
	ROM Force	Fail Force	Fail Distance	ROM Force	Fail Force	Fail Distance
Specimen 1	298	298	0.539	14	23	3.469
Specimen 2	290	290	0.574	15	25	3.656
Specimen 3	314	314	0.789	15	22	3.499
Specimen 4	286	286	0.635	16	26	3.438
Specimen 5	310	310	0.295	15	24	3.938

Note: ROM: theoretical range of motion of prototype joint; Fail: Failure

Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under tensile stress at zero degrees to the long axis of the linkage. Force data reported in pounds-force (lbf) and deflection data reported in inches (in).

As it pertains to maximum force transfer within the limits of the theoretical range of motion for control specimen; a mean of 300 pounds-force (*SD* = 12.2), with a 95% confidence interval of [289, 310] was observed ($n = 5$). For prototype specimen; a mean of 15.0 pounds-force (*SD* = 0.71), with a 95% confidence interval of [14.4, 15.6] was observed ($n = 5$). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 52.08$. Significant differences in data for the control ($M = 300$, $SD = 12.2$) and prototype ($M = 15.0$, $SD = 0.71$) specimen were observed: $t(4) = 52.08$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, see Figure 16.

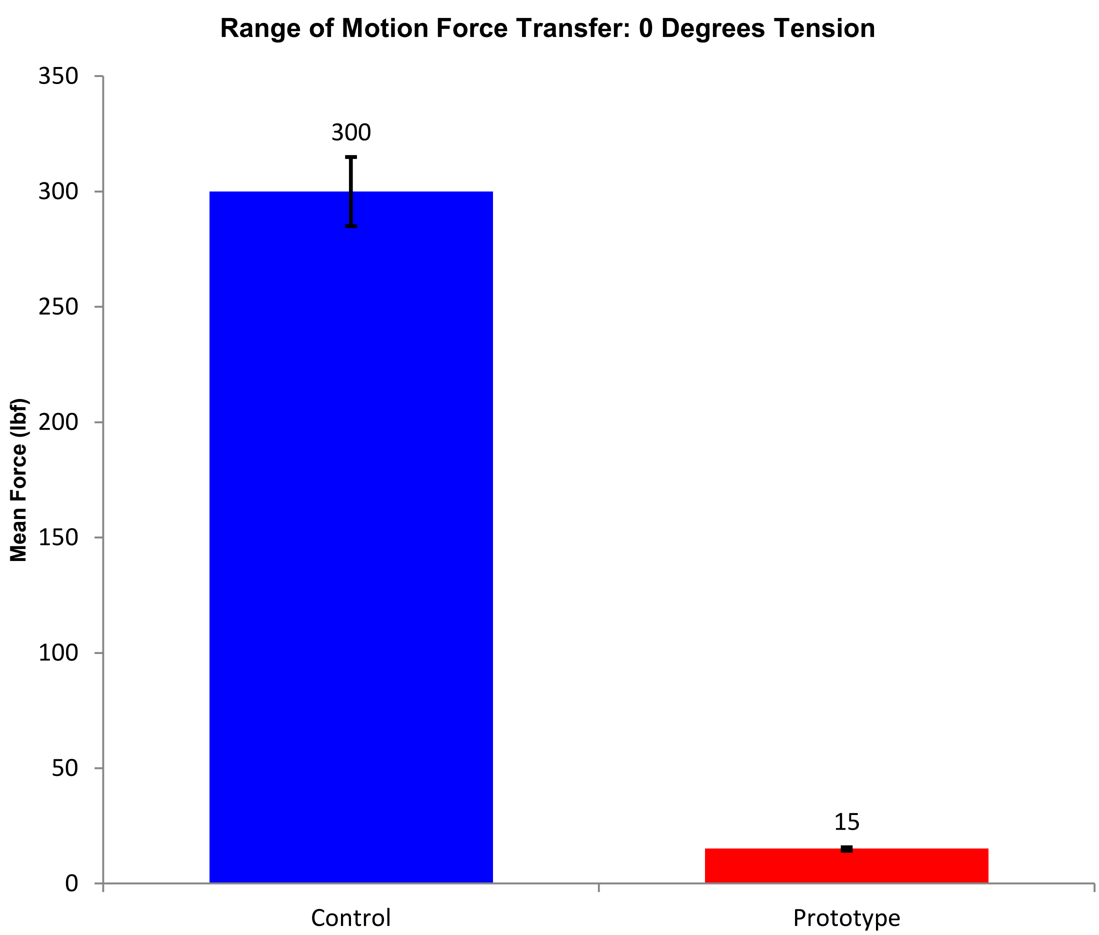

Figure 16. Significant differences in data for the control ($M = 300$, $SD = 12.2$) and prototype ($M = 15.0$, $SD = 0.71$) specimen were observed: $t(4) = 52.08$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to maximum force transfer at failure for control specimen; a mean of 300 pounds-force (*SD* = 12.2), with a 95% confidence interval of [289, 310] was observed ($n = 5$). For prototype specimen; a mean of 24 pounds-force (*SD* = 1.58), with a 95% confidence interval of [22.6, 25.4] was observed ($n = 5$). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 50.10$. Significant differences in data for the control ($M = 300$, $SD = 12.2$) and prototype ($M = 24$, $SD = 1.58$) specimen were observed: $t(4) = 50.10$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, Figure 17.

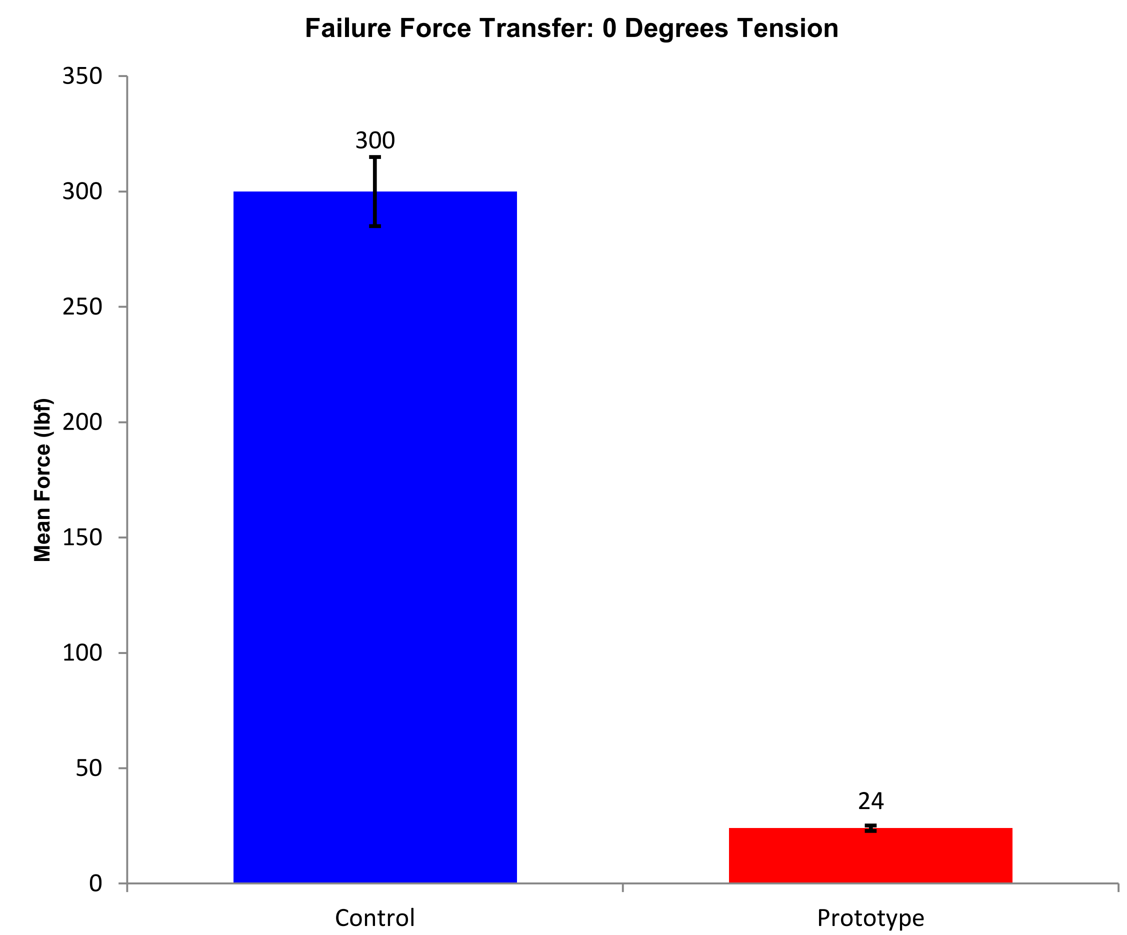

Figure 17. Significant differences in data for the control ($M = 300$, $SD = 12.2$) and prototype ($M = 24$, $SD = 1.58$) specimen were observed: $t(4) = 50.10$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to deflection at failure for control specimen; a mean of 0.57 inches (SD = 0.18), with a 95% confidence interval of [0.41, 0.72] was observed (n = 5). For prototype specimen; a mean of 3.60 inches (SD = 0.21), with a 95% confidence interval of [3.42, 3.78] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 24.78$. Significant differences in data for the control (M = 0.57, SD = 0.18) and prototype (M = 3.60, SD = 0.21) specimen were observed: $t(4) = 24.78$, $p < 0.001$; one-tailed t distribution critical value = 1.86, two-tailed t distribution critical value = 2.31, Figure 18.

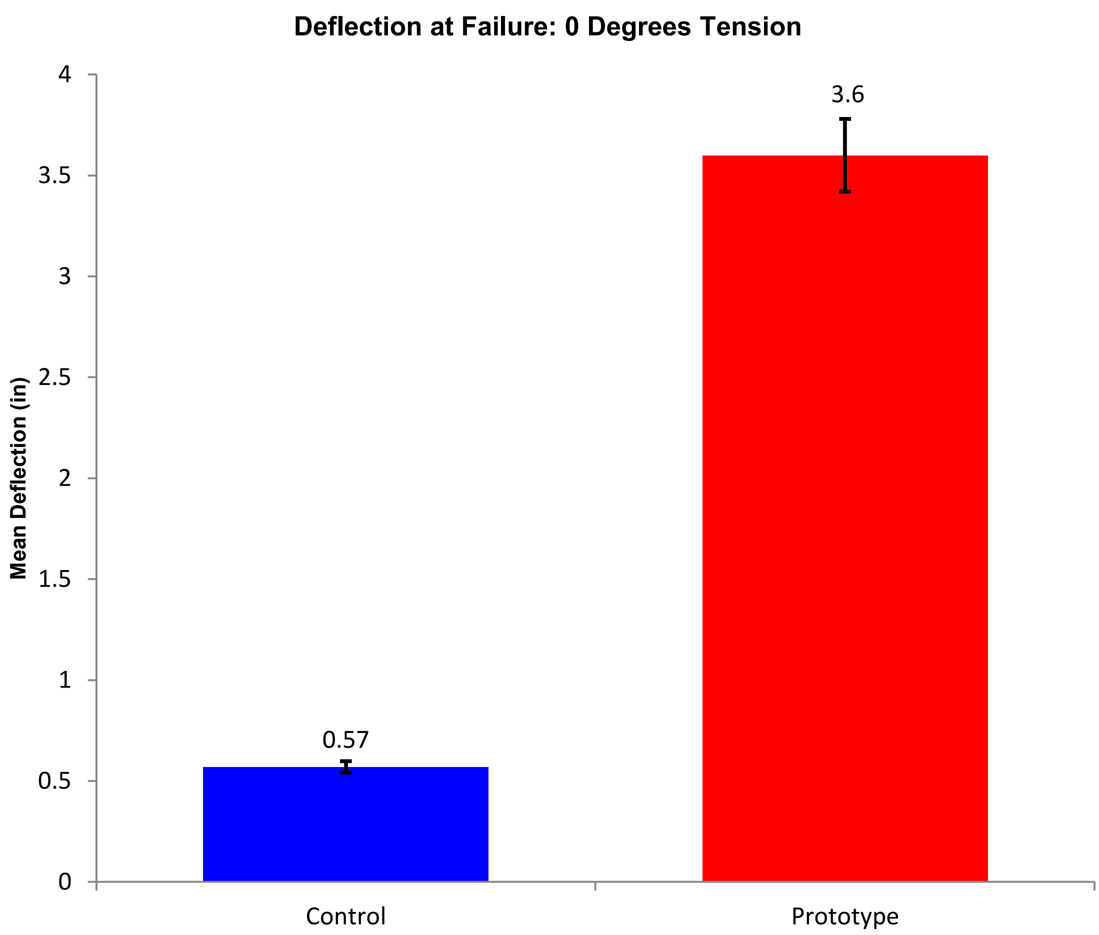

Figure 18. Significant differences in data for the control (M = 0.57, SD = 0.18) and prototype (M = 3.60, SD = 0.21) specimen were observed: $t(4) = -24.78$, $p < 0.001$; one-tailed t distribution critical value = 1.86, two-tailed critical value = 2.31.

Tension at 45°

Force transfer testing in tension at 45 degrees to the long axis of the mechanical linkage for control and prototype specimen (N = 10) was carried out with custom-fabricated fixtures. Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under forty-five degree tensile stress is located in Table 5.

Table 5: Tension at 45° (Degrees)

	Control			Prototype		
	ROM Force	Fail Force	Fail Distance	ROM Force	Fail Force	Fail Distance
Specimen 1	301	303	1.081	15	23	3.469
Specimen 2	293	294	1.029	15	25	3.656
Specimen 3	291	291	0.939	14	22	3.499
Specimen 4	300	303	1.079	14	26	3.438
Specimen 5	288	288	1.006	14	24	3.938

Note: ROM: theoretical range of motion of prototype joint; Fail: Failure

Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under tensile stress at forty-five degrees to the long axis of the linkage. Force data reported in pounds-force (lbf) and deflection data reported in inches (in).

As it pertains to maximum force transfer within the limits of the theoretical range of motion for control specimen; a mean of 295 pounds-force (SD = 5.68), with a 95% confidence interval of [290, 300] was observed (n = 5). For prototype specimen; a mean of 14.4 pounds-force (SD = 0.55), with a 95% confidence interval of [13.9, 14.9] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 109.73$. Significant differences in data for the control (M = 295, SD = 5.6) and prototype (M = 14.4, SD = 0.55) specimen were observed: $t(4) = 109.73$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, see Figure 19.

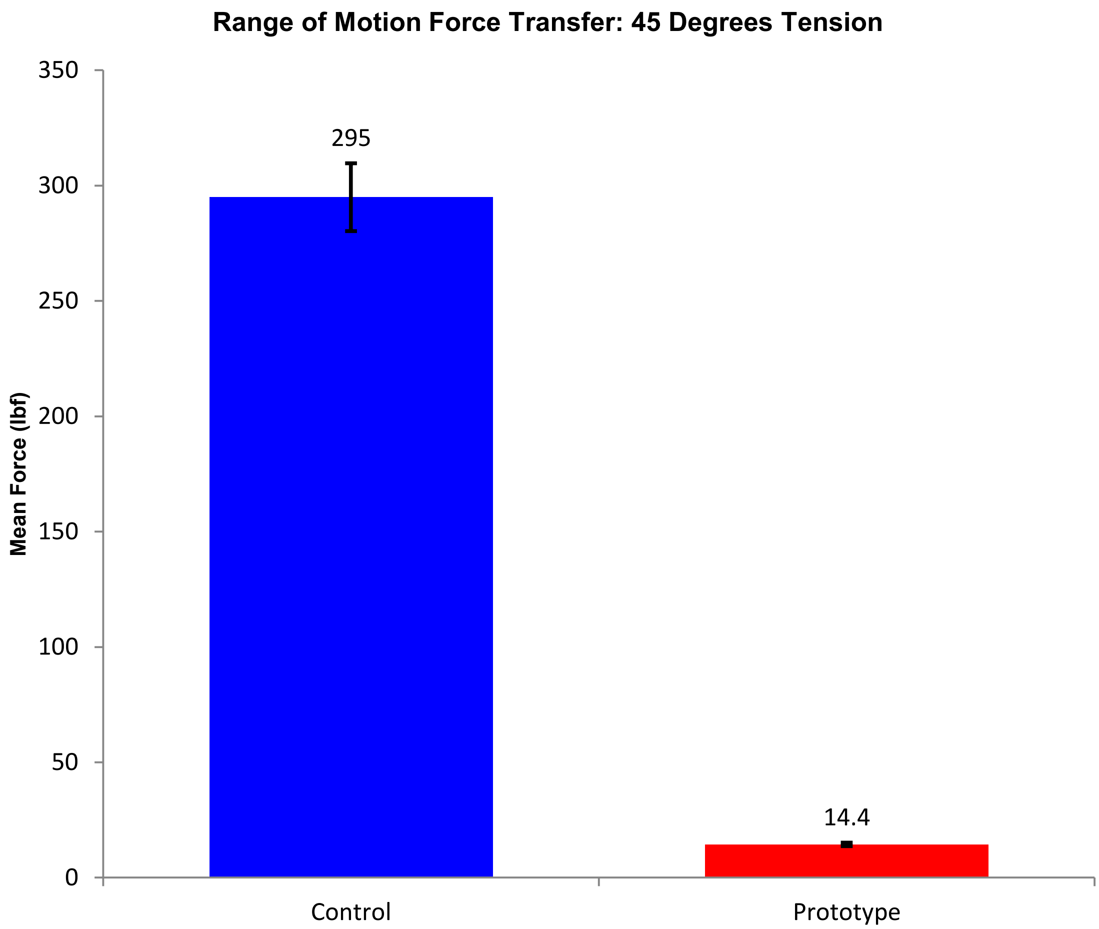

Figure 19. Significant differences in data for the control (M = 295, SD = 5.6) and prototype (M = 14.4, SD = 0.55) specimen were observed: $t(4) = 109.73$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to maximum force transfer at failure for control specimen; a mean of 296 pounds-force (SD = 6.91), with a 95% confidence interval of [290, 302] was observed (n = 5). For prototype specimen; a mean of 24 pounds-force (SD = 1.58), with a 95% confidence interval of [22.6, 25.4] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 85.78$. Significant differences in data for the control (M = 296, SD = 6.91) and prototype (M = 24, SD = 1.58) specimen were observed: $t(4) = 85.78$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, Figure 20.

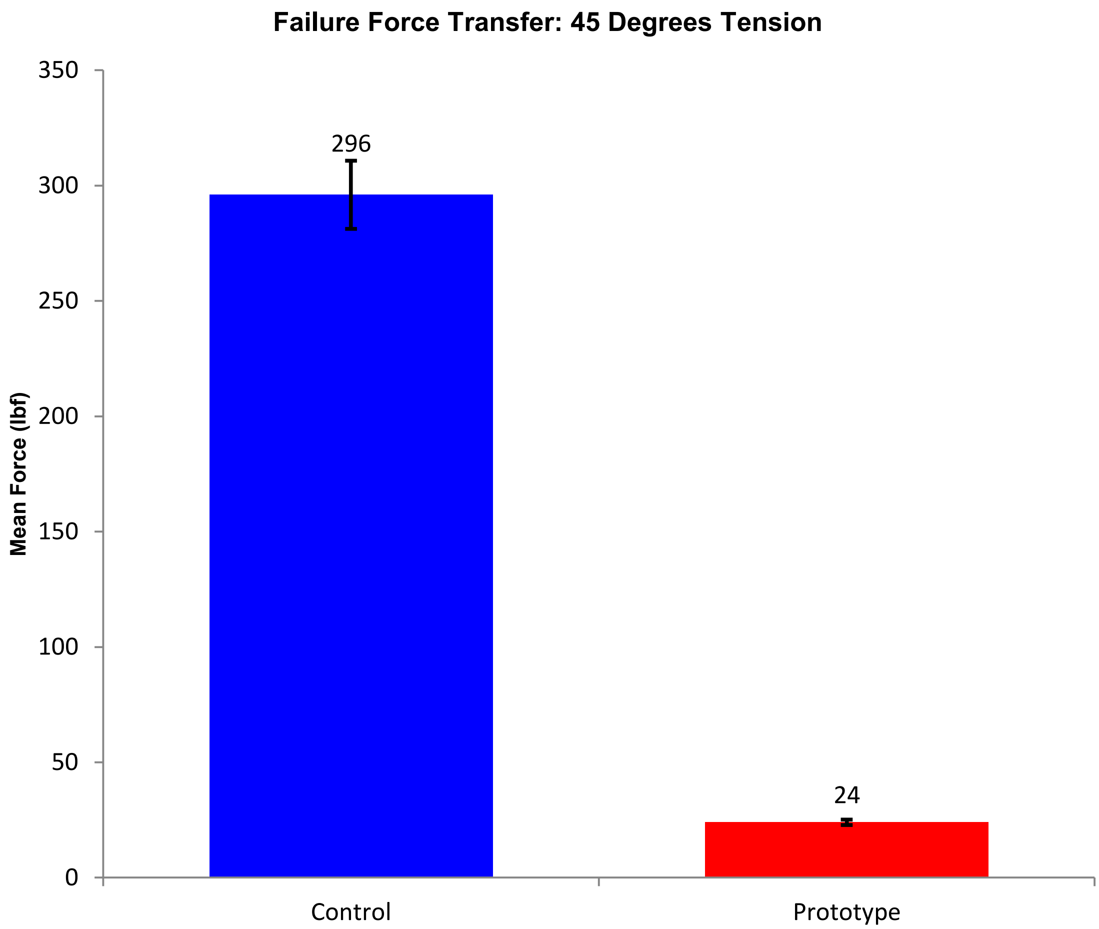

Figure 20. Significant differences in data for the control (M = 296, SD = 6.91) and prototype (M = 24, SD = 1.58) specimen were observed: $t(4) = 85.78$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to deflection at failure for control specimen; a mean of 1.03 inches (SD = 0.06), with a 95% confidence interval of [0.98, 1.08] was observed (n = 5). For prototype specimen; a mean of 3.60 inches (SD = 0.21), with a 95% confidence interval of [3.42, 3.78] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 26.77$. Significant differences in data for the control (M = 1.03, SD = 0.06) and prototype (M = 3.60, SD = 0.21) specimen were observed: $t(4) = 26.77$, $p < 0.001$; one-tailed t distribution critical value = 2.02, two-tailed t distribution critical value = 2.57, Figure 21.

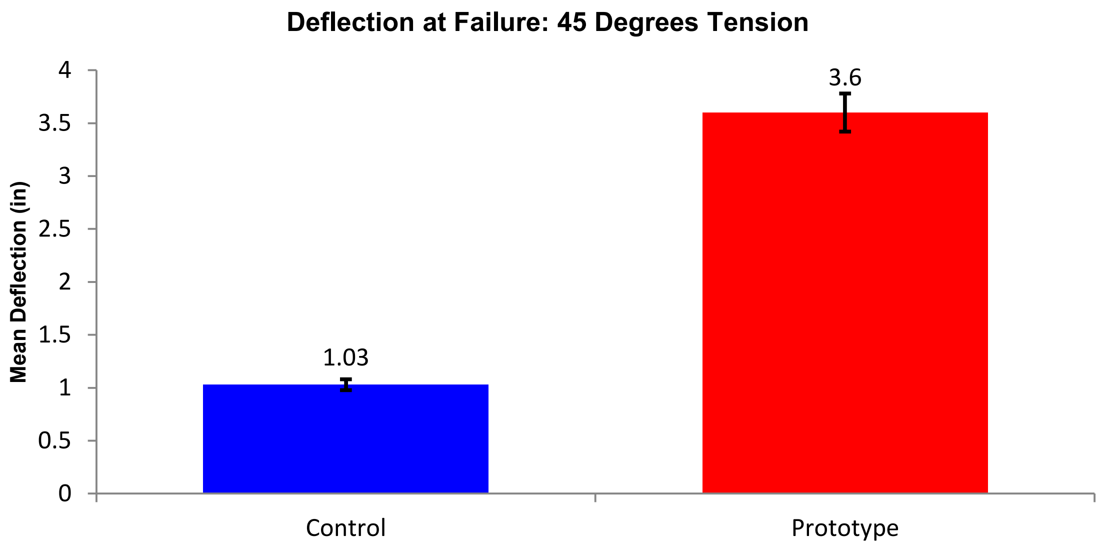

Figure 21. Significant differences in data for the control (M = 1.03, SD = 0.06) and prototype (M = 3.60, SD = 0.21) specimen were observed: $t(4) = -26.77$, $p < 0.001$; one-tailed t distribution critical value = 2.02, two-tailed critical value = 2.57.

Tension at 90°

Force transfer testing in tension at 90 degrees to the long axis of the mechanical linkage for control and prototype specimen (N = 10) was carried out with custom-fabricated fixtures. Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under ninety degree tensile stress is located in Table 6.

Table 6: Tension at 90^{o} (Degrees)

	Control			Prototype		
	ROM Force	Fail Force	Fail Distance	ROM Force	Fail Force	Fail Distance
Specimen 1	234	254	1.227	13	23	3.469
Specimen 2	256	263	1.134	13	25	3.656
Specimen 3	249	252	1.096	14	22	3.499
Specimen 4	243	279	1.253	13	26	3.438
Specimen 5	247	265	1.192	14	24	3.938

Note: ROM: theoretical range of motion of prototype joint; Fail: Failure

Raw data for force transfer from the facemask connector to the helmet connector and deflection of the receptacle at failure under tensile stress at ninety degrees to the long axis of the linkage. Force data reported in pounds-force (lbf) and deflection data reported in inches (in).

As it pertains to maximum force transfer within the limits of the theoretical range of motion for control specimen; a mean of 246 pounds-force (*SD* = 8.11), with a 95% confidence interval of [239, 253] was observed (*n* = 5). For prototype specimen; a mean of 13.4 pounds-force (*SD* = 0.55), with a 95% confidence interval of [1.12, 1.24] was observed (*n* = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 63.97$. Significant differences in data for the control ($M = 246$, $SD = 8.11$) and prototype ($M = 13.4$, $SD = 0.55$) specimen were observed: $t(4) = 63.97$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, see Figure 22.

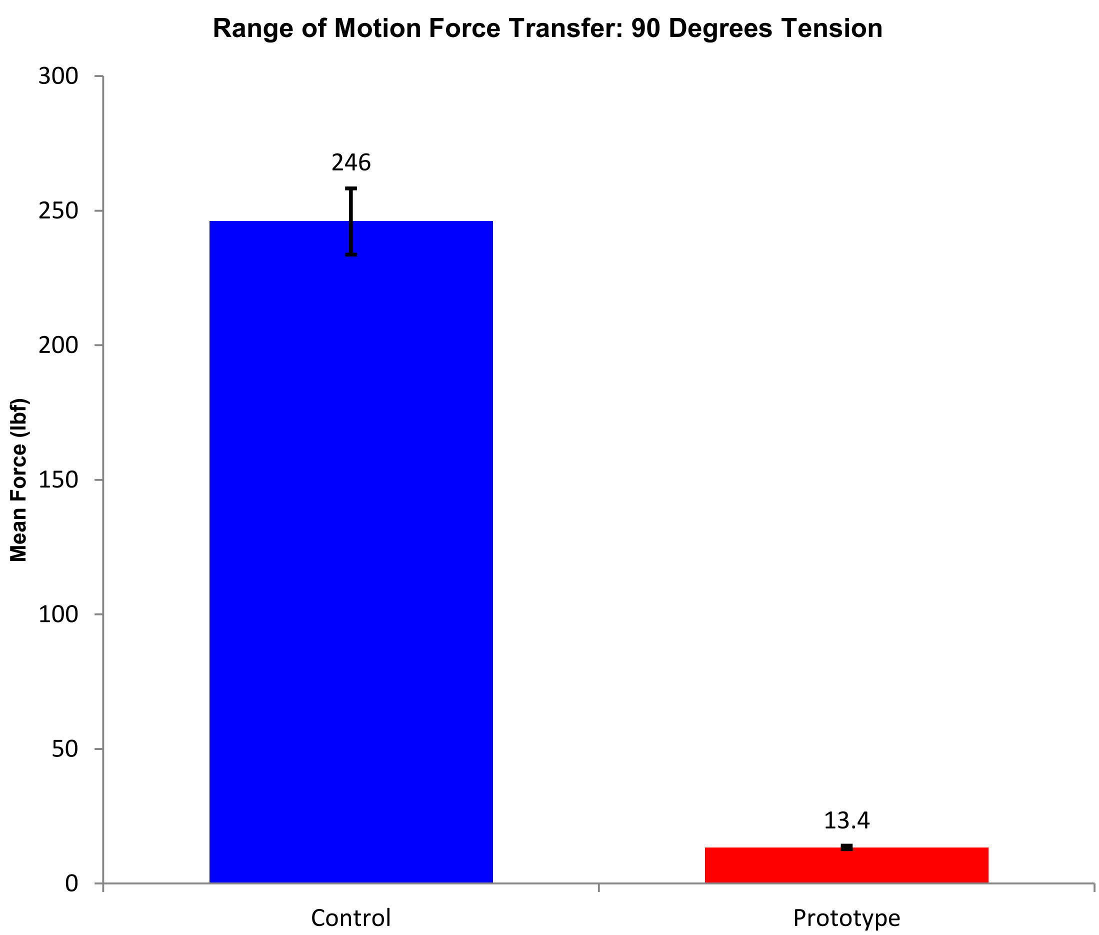

Figure 22. Significant differences in data for the control ($M = 246$, $SD = 8.11$) and prototype ($M = 13.4$, $SD = 0.55$) specimen were observed: $t(4) = 63.97$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to maximum force transfer at failure for control specimen; a mean of 263 pounds-force (*SD* = 10.7), with a 95% confidence interval of [253, 272] was observed ($n = 5$). For prototype specimen; a mean of 24 pounds-force (*SD* = 1.58), with a 95% confidence interval of [22.6, 25.4] was observed ($n = 5$). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 49.16$. Significant differences in data for the control ($M = 263$, $SD = 10.7$) and prototype ($M = 24$, $SD = 1.58$) specimen were observed: $t(4) = 49.16$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed t distribution critical value = 2.78, Figure 23.

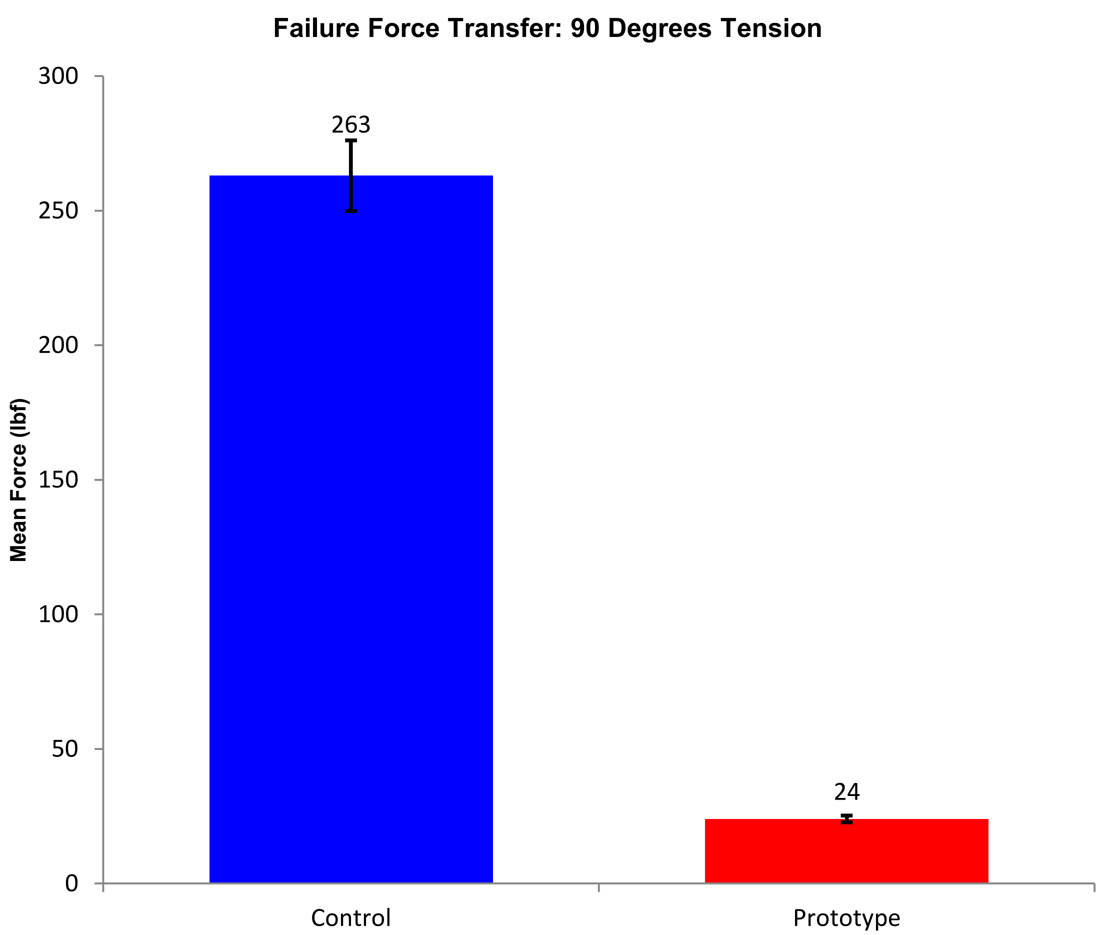

Figure 23. . Significant differences in data for the control ($M = 263$, $SD = 10.7$) and prototype ($M = 24$, $SD = 1.58$) specimen were observed: $t(4) = 49.16$, $p < 0.001$; one-tailed t distribution critical value = 2.13, two-tailed critical value = 2.78.

As it pertains to deflection at failure for control specimen; a mean of 1.18 inches (*SD* = 0.06), with a 95% confidence interval of [1.12, 1.24] was observed (n = 5). For prototype specimen; a mean of 3.60 inches (*SD* = 0.21), with a 95% confidence interval of [3.42, 3.78] was observed (n = 5). A t-test assuming unequal variance to compare means for independent samples, alpha equal to 0.05, revealed $t(4) = 24.97$. Significant differences in data for the control (M = 1.18, *SD* = 0.06) and prototype (M = 3.60, *SD* = 0.21) specimen were observed: $t(4) = 24.97$, $p < 0.001$; one-tailed t distribution critical value = 2.02, two-tailed t distribution critical value = 2.57, Figure 24.

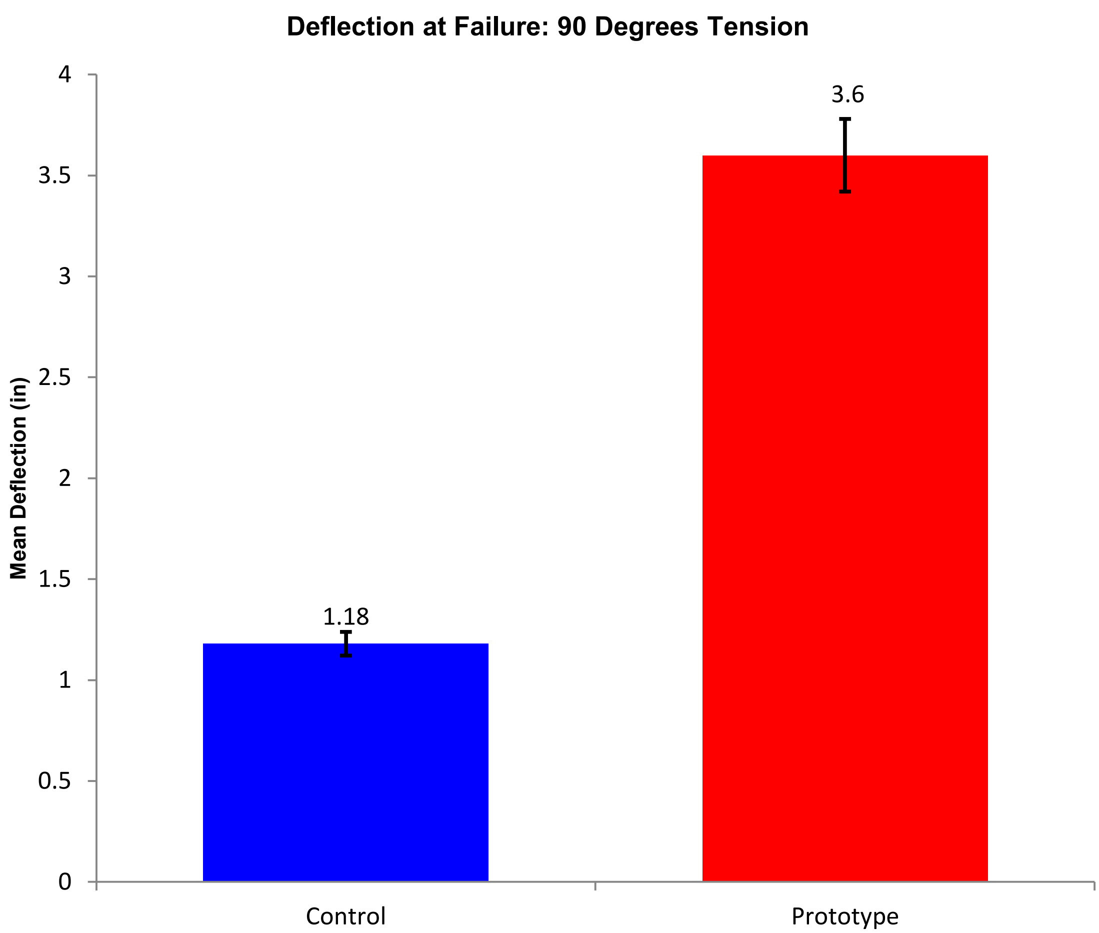

Figure 24. Significant differences in data for the control (M = 1.18, *SD* = 0.06) and prototype (M = 3.60, *SD* = 0.21) specimen were observed: $t(4) = -24.97$, $p < 0.001$; one-tailed t distribution critical value = 2.02, two-tailed critical value = 2.57.

CHAPTER 5: DISCUSSION AND CONCLUSIONS

Research Question 1: Hypothesis Assessment

Is it possible to design a novel mechanical linkage of reasonable simplicity that provides 360° of freedom in motion capable of decreasing force transfer from the facemask to the protective helmet when loading occurs? To adequately answer this question, we must evaluate the null and alternate hypotheses with regard to the third condition, linkage deflection at failure ($M_{C0} = M_{P0}$ or $M_{C0} \neq M_{P0}$; $M_{C45} = M_{P45}$ or $M_{C45} \neq M_{P45}$; $M_{C90} = M_{P90}$ or $M_{C90} \neq M_{P90}$), individually, and subsequently interpret the findings as a whole, either in acceptance or rejection of the null and alternate hypotheses.

For each condition measured under both compressive and tensile force, statistically significant differences between the control and prototype groups were observed. Findings for the deflection at failure of the mechanical linkage under compressive force at zero degrees $t(6) = 20.55$ ($p < 0.001$), at forty-five degrees $t(6)$ 26.85 ($p < 0.001$) and at ninety degrees $t(5) = 24.97$ ($p < 0.001$); in combination with findings under tensile force at zero degrees $t(8) = 24.78$ ($p < 0.001$), at forty-five degrees $t(5)$ 26.77 ($p < 0.001$) and at ninety degrees $t(5) = 24.97$ ($p < 0.001$), indicate an increased flexibility of the prototype mechanical linkage. Manual manipulation, as Figure 25 demonstrates photographically, reveals 360° of freedom in motion of the prototype mechanical linkage. In addition, the materials used to construct the prototype linkage, as described above, are readily available and of reasonable cost.

The preceding allows us to reject the null hypothesis and accept the alternate hypothesis, and state that, designing a novel mechanical linkage of reasonable simplicity that provides 360° of freedom in motion is possible.

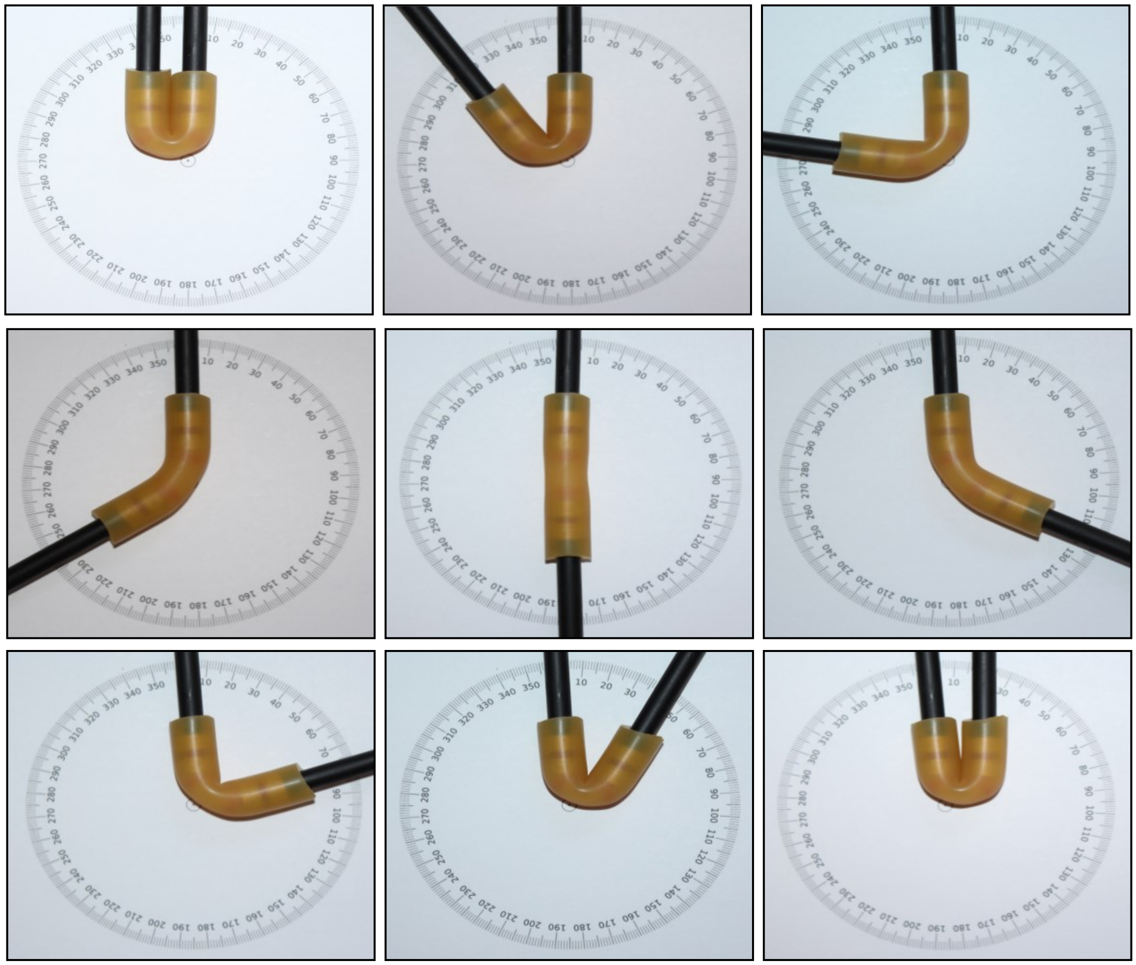

Figure 25. Prototype Mechanical Linkage Freedom of Motion

Research Question 2: Hypothesis Assessment

Can significant decreases in force transfer be obtained when compressive (frontal impact) forces are applied to the prototype mechanical linkage? To adequately answer this question, we must evaluate the null and alternate hypotheses with regard to the first and second conditions, maximum force transfer at the theoretic range of motion and at failure ($M_{C0} = M_{P0}$ or $M_{C0} \neq M_{P0}$; $M_{C45} = M_{P45}$ or $M_{C45} \neq M_{P45}$; $M_{C90} = M_{P90}$ or $M_{C90} \neq M_{P90}$), individually, and subsequently interpret the findings as a whole, either in acceptance or rejection of the null and alternate hypotheses.

For each condition measured under compressive force, statistically significant differences between the control and prototype groups were observed. Findings for maximum force transfer within the limits of the theoretical range of motion at zero degrees $t(4) = 44.51$ ($p < 0.001$), at forty-five degrees $t(4) = 31.17$ ($p < 0.001$) and at ninety degrees $t(4) = 63.97$ ($p < 0.001$); indicate a significant decrease in the force transfer from the facemask connector to helmet connector. Findings for maximum force transfer at failure at zero degrees $t(4) = 14.27$ ($p < 0.001$), at forty-five degrees $t(4) = 29.13$ ($p < 0.001$) and at ninety degrees $t(4) = 49.16$ ($p < 0.001$); also indicate a significant decrease in the force transfer from the facemask connector to helmet connector.

The preceding allows us to reject the null hypothesis and accept the alternate hypothesis, and state that, the prototype mechanical linkage decreased measured force transfer from the facemask component to helmet component when compressive force was applied at 0^0, 45^0, and 90^0. As a result, it can be reasonably inferred that the significant decrease in force transfer from the facemask connector to the helmet connector has the potential to prolong the duration of a collision while reducing the total momentum transferred to the head.

Research Question 3: Hypothesis Assessment

Can significant decreases in force transfer be obtained when tensile (pulling) forces are applied to the prototype mechanical linkage? To adequately answer this question, we must evaluate the null and alternate hypotheses with regard to the first and second conditions, maximum force transfer at the theoretic range of motion and at failure ($M_{C0} = M_{P0}$ or $M_{C0} \neq M_{P0}$; $M_{C45} = M_{P45}$ or $M_{C45} \neq M_{P45}$; $M_{C90} = M_{P90}$ or $M_{C90} \neq M_{P90}$), individually, and subsequently interpret the findings as a whole, either in acceptance or rejection of the null and alternate hypotheses.

For each condition measured under tensile force, statistically significant differences between the control and prototype groups were observed. Findings for maximum force transfer within the limits of the theoretical range of motion at zero degrees $t(4) = 52.08$ ($p < 0.001$), at forty-five degrees $t(4) = 52.08$ ($p < 0.001$) and at ninety degrees $t(4) = 52.08$ ($p < 0.001$); indicate a significant decrease in the force transfer from the facemask connector to helmet connector. Findings for maximum force transfer at failure at zero degrees $t(4) = 50.10$ ($p < 0.001$), at forty-five degrees $t(4) = 85.78$ ($p < 0.001$) and at ninety degrees $t(4) = 49.16$ ($p < 0.001$); also indicate a significant decrease in the force transfer from the facemask connector to helmet connector.

The preceding allows us to reject the null hypothesis and accept the alternate hypothesis, and state that, the prototype mechanical linkage decreased measured force transfer from the facemask component to helmet component when tensile force was applied at 0^0, 45^0, and 90^0. As a result, it can be reasonably inferred that the significant decrease in force transfer from the facemask connector to the helmet connector has the potential to prolong the duration of a collision while reducing the total momentum transferred to the head.

Significance to Sports Medicine

Helmets decrease the potential for traumatic brain injury following a collision by reducing the acceleration of the head upon impact; by this means decreasing both the brain-skull collision, as well as the sudden deceleration induced axonal injury (Daneshvar et al, 2011). Energy absorbing materials within helmets, which act by compressing to absorb force during a collision and slowly returning to their original shape, prolong the duration of the collision, while reducing the total momentum transferred to the head (Daneshvar et al, 2011). Incorporation of the prototype mechanical linkage designed as part of this study has the potential to augment ongoing advances in helmet technology. Theoretically, the prototype mechanical linkage would act to further prolong the duration of the injurious event, reducing momentum transfer and ultimately the acceleration of the head; either upon frontal impact or when pulled upon forcefully.

Nevertheless, one area of concern with regard to the performance of the prototype mechanical linkage lies in the fact that a relatively low force was required to incite failure. During gameplay, early facemask failure could leave an athlete exposed to additional and unnecessary injury. The force requirement to incite failure of the prototype mechanical linkage was approximately ten percent that of the control for all angulations in both tension and compression, with a mean of 24 pounds-force. Certainly, the facemasks of athletes participating in sports that involve extensive personal contact, will endure forces that exceed the 24 pounds-force threshold for failure. Fortunately, under the static experimental conditions described, the force to incite failure had to be sustained for an average of 164 seconds, or 2.73 minutes. Loading of this duration is highly unlikely to occur during normal gameplay. Therefore, as it pertains to static loading, it is assumed that these numbers are of little significance.

As previously stated, according to the National Spinal Cord Injury Statistical Center, approximately 12,000 new cases of spinal cord injuries occur each year, with sports-related events causing approximately 7.6% of the injuries (Zahir & Ludwig, 2010).

Football is associated with the largest number of overall catastrophic cervical spine injuries according to the National Center for Catastrophic Sports Injury Research (Boden, Tacchetti, Cantu, Knowles, & Mueller, 2006). In relation, high-school and collegiate athletes endure an average of 7.23 direct catastrophic head injuries per year (Boden, Tacchetti, Cantu, Knowles, & Mueller, 2007) and nearly 85% of all football-related fatalities, between 1945 and 1994, resulted from head and cervical spine injuries (Zahir & Ludwig, 2010).

Such events are often life-altering events for not only the individual involved, but also their families and friends, with far-reaching implications of unfathomable magnitude. By that measure, any improvement, no matter how miniscule, that could be afforded by the prototype mechanical linkage, as it pertains to the aforementioned population data is of significance.

Study Limitations

Possible methodological and researcher limitations to this project include, but may not be limited to, the lack of prior research on the specified topic, a lack of available control data, longitudinal effects and inadequate sample size. Research on the specific problem that this project aimed to evaluate is not readily available in the literature. As a result, the study was designed in a theoretical and exploratory fashion, with no well-known baseline for comparison. In relation, the lack of available data for use as a viable control meant extensive planning and jig fabrication were necessary to establish said control. Consequently, important research man-hours were lost that could have otherwise been dedicated to testing the prototype mechanical linkage more extensively. The longitudinal time constraints of the Orthodontic Certificate/Master of Oral Biology program, in combination with scant financial resources, led to an unavoidable limitation of the sample size, inevitably decreasing the power of the findings.

Recommendations for Further Research

Evaluation of Cranial Acceleration

At impact, the head is likely to encounter both linear and rotational accelerations, damaging neural and vascular elements of the central nervous system (Barth, Freeman, Broshek & Varney, 2001). To evaluate cranial acceleration, current data supports that an accelerometer placed intra-orally, via mouth-guard, measures acceleration more accurately than an accelerometer placed on the helmet (Higgins, Halstead, Synder-Mackler, & Barlow, 2007). The methodology of a future study should follow the accepted method of impact testing using biofidelic headforms, endorsed by the National Operating Committee on Standards for Athletic Equipment in the impact testing of football, hockey, baseball, and lacrosse helmets. The objective should be to evaluate cranial acceleration when impact is made with the facemask of a helmet and face-mask system fitted with the prototype mechanical linkage designed as part of this project compared to a traditional helmet and face-mask system.

Evaluation of Facemask Removal

For players whom experience suspected cervical spinal injuries, it is the current recommendation to remove the facemask instead of the helmet (Banarjee & Palumbo, 2004). Techniques of facemask removal, including cutting the loop straps with various tools, and removing the loop straps with a cordless screwdriver, have been investigated (Swartz, Belmore, Decoster & Armstrong, 2010). The objective of a future study should be similar to that conducted by Swartz, Belmore, Decoster & Armstrong in 2010, comparing the efficiency of face-mask removal with regard to success rates, time, head motion, and difficulty between a helmet fitted with the prototype mechanical linkage designed as part of this project and traditional helmet and face-mask system.

Evaluation of Alternative Materials

Contemporary advances in materials science offer a seemingly limitless ability to customize components of the prototype mechanical linkage to assume any combination of physical properties desirable. Companies, such as C & M Rubber Co. claim to be capable of producing custom compounds that can tolerate wide ranges of temperatures, tear resistance, and compression set.

In future studies, different materials for the receptacle component of the prototype mechanical linkage should be tested to evaluate the desired combination of physical properties, including: resilience, tensile strength, elongation, shear strength, coefficient of friction, impact resistance, resistance to abrasion, and resistance to tear; until an optimal receptacle material is found or formulated.

Conclusion

When compared to currently available designs, the prototype mechanical linkage designed and tested as part of this project is of reasonable simplicity, displays increased flexibility and provides 360° of freedom in motion. Under compressive and tensile forces, force transfer from the facemask component to helmet component was decreased significantly. As a result, it can be reasonably inferred that the significant decrease in force transfer from the facemask connector to the helmet connector has the potential to prolong the duration of a collision while reducing the total momentum transferred to the head.

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