Comparing Equestrian Helmets with and without Rotational Technology using an Equestrian Concussive Specific Helmet Test Protocol.

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Abstract

Horse riding is a popular global activity involving a wide range of sporting events including dressage, endurance riding, eventing, show jumping, horse racing and rodeo. Unfortunately, horse riding and equestrian sporting events, report a high prevalence of concussion. The most common mechanism for brain injury in equestrian events involve high levels of linear and rotational acceleration during head impacts when falling from a horse. These accelerations create injurious brain tissue strain. While both linear and rotational accelerations occur during head impacts, the rotational components of acceleration are closely linked to brain tissue strain. To reduce brain strain, helmet technologies have been developed with the aim to reduce head rotational accelerations during an impact.

The most common rotational managing technology, multi-directional impact protection system (MIPS), employs a low friction layer to reduce the amount of rotational acceleration sustained by the brain during head impacts. MIPS tests equestrian helmets using a monorail drop rig with a 45-degree steel anvil covered in 80 grit sandpaper at 6.2m/s. The surface experiencing impact in the MIPS test method is a very low compliant surface (steel). It is impacted at a velocity of 6m/s, and an anvil angle of 45-degrees. In contrast, most impacts in equestrian involve high compliant material such as sand or turf with an average impact velocity is 9m/s, and the average angle of impact of 27 degrees. The proposed rotational testing method employed by MIPS may not fully represent the most common accidents involving equestrian events.

The objective of this research was to evaluate the effectiveness of a helmet with rotational technology to reduce linear and rotational acceleration, rotational velocity, and maximum principal strain (MPS) in equestrian helmets.

An equestrian specific test protocol was developed using the common impact conditions for concussive events for equestrian riders. Nine m/s impact velocity, with an angle of 26.5 degrees to the horizontal axis, and an anvil compliance consisting of 66mm of 602 vinyl nitrile foam with synthetic grass to represent turf impacts was reported as the most common impact characteristics. Using a Rail Guided Launcher, a helmeted Hybrid III headform was launched and impacted a low and high compliance anvil using the defined velocity and angle parameters. Two equestrian helmet types were impacted, a conventional helmet with no rotational technology and the same helmet model with rotational technology. The impact locations tested included front, side, and rear boss, as these were the most common impact locations reported for concussive events in equestrian. Linear and rotational acceleration and rotational velocity were measured using a DTS SLICE sensor installed inside the headform. The linear and rotational acceleration curves were then used as input to the University College Dublin Brain Trauma Model (V2.0) to calculate MPS. Statistical analysis included four t-tests, two 2x2x3 ANOVA's with 8 pairwise Tukey post-hoc test, significance set to $\alpha=0.05$.

The results were not uniform across impact locations and anvil compliances, the rear boss impact location in helmets with rotational technology revealed significantly lower rotational accelerations and rotational velocity. The results revealed helmets with rotational technology should be designed to perform under these high energy conditions. If the rotational technology was designed with these considerations, it would be possible to investigate the potential of rotational technologies to decrease dynamic head response and the brain tissue strain.

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Introduction

Equestrian sports are popular around the world, particularly in the United Kingdom, France, and Ireland (Forero Rueda & Gilchrist, 2010). Equestrians of all levels of ability are vulnerable to sustaining injuries (Turner et al, 2002), with the most common catastrophic injury events involving the head (Havlik, 2010; Ball et al, 2007, Clark et al, 2020; Connor et al, 2019; Walker et al, 2000; McCrory & Turner, 2005). Research conducted by Winkler and colleagues (2016) reported that from 2003-2012, the highest reported percentage of sport related traumatic brain injuries (TBI) were in equestrian sports when compared to other sports, including football, rugby, and skiing. Equestrian participation comes with implicit risk as the rider's head can be positioned 2-3 meters above the ground and traveling at speeds of 9-18m/s (Havlik, 2010; Forero Rueda & Gilchrist, 2010). As with many other sports, helmet use is common in racing and eventing equestrian disciplines to reduce the risk of traumatic brain injury. Helmets are primarily designed to protect against catastrophic injury using linear acceleration managing materials. Concussions are associated with rotational accelerations (Forero Rueda, Cui, and Gilchrist, 2011; Gennarelli et al, 1972, Kleiven, 2006**)**, indicating rotational acceleration managing technology materials are needed. Head injuries, particularly rates of concussions among equestrians remain high (Connor & Clark et al, 2019), suggesting there is opportunity to improve the protective capacity of helmets (Stanfill et al, 2020; Clark, Connor, et al, 2020; Clark, Adanty et al, 2020; Forero Rueda, Cui et al, 2010).

Concussions are caused by a direct or indirect impact to the head and neck, presenting a wide range of clinical symptoms including physical, behavioural, cognitive impairment, and sleep disturbances (McCrory et al, 2017; Zhang et al, 2004). Real-world impact reconstructions of 50 events performed by Clark, Adanty and colleagues (2020) reported that in equestrian concussive events, concussive accidents primarily occur when an athlete falls obliquely to a compliant surface such as turf or sand at an average velocity of 9m/s. Given the speed of the sport and the height of the fall, the brain is vulnerable to high magnitudes of strain upon impact. While both linear and rotational accelerations occur from impacts to the head, rotational components are more closely linked to brain tissue strain and concussion. This is due to the mechanism by which rotational accelerations cause large magnitude strains to the brain given its low shear modulus (Post & Hoshizaki, 2015). It is important that helmets are designed to mitigate both linear and rotational acceleration to better protect the brain from structural damage that often results in brain injury.

Multi-directional impact protection system (MIPS) is a low friction layer added to helmets to reduce the amount of rotational energy the brain sustains; however, it is unknown how rotational technologies in equestrian helmets influence the dynamic head and brain tissue response in a real-world testing protocol. The purpose of this study is to determine if rotational technology used in equestrian helmets reduce the dynamic head response and the magnitude of strain the brain sustains under impact conditions using an equestrian specific protocol.

Significance

Current helmets with rotation managing technology are not tested using real-world impact protocols. This research will provide an understanding of how helmets with rotational technology perform using an equestrian specific test protocol. The findings from this research will provide information to guide future helmet technology innovation to reduce the risk of concussive brain injury in equestrian riders.

Literature Review

Dependent Variables for Brain Injury

Linear acceleration is associated with brain motion causing intracranial pressure waves and is used as the primary measure to evaluate helmet performance for the risk of brain injury (Gurdjian et al, 1958, 1963; Gurdjian, 1975; Thomas et al, 1966, 1967). Linear acceleration is also used to predict the risk of skull fracture, with most current sport helmet standards using peak linear acceleration as the variable for pass or fail criteria (Post & Hoshizaki, 2012; Hoshizaki & Brien, 2004). Head impacts in sport seldom result in exclusively linear acceleration and therefore, it is important to consider both linear and rotational accelerations.

Focal point shear strains that cause brain injury are primarily the result of rotational acceleration of brain tissue during an impact (Bradshaw and Morfey, 2001; Hardy et al, 1994). Rotational acceleration creates diffuse shearing of brain tissue resulting in focal strains due to the differing densities of grey and white matter (Ommaya and Gennarelli, 1974; Yoganandan et al, 2008). Brain tissue is sensitive to rotational motion due to the mechanism by which rotational accelerations create large strain magnitudes given its low shear modulus (Post & Hoshizaki, 2015). The brain has a high resistance to compressive forces that are associated with translation accelerations created by linear forces (Ommaya, 1968; Ommaya & Hirsch, 1971). Rotational acceleration should be considered in testing methods to improve the capacity of helmets to protect against concussions (Forero Rueda, Cui, and Gilchrist, 2011; Greenwald et al, 2008).

Rotational velocity is correlated with rotational acceleration, brain tissue response and concussions and is used as a surrogate for measuring the risk of brain injury (Kleiven, 2007; Gennarelli, 1981; Forero Rueda & Gilchrist, 2010).

Maximum principal strain (MPS) is a multidirectional measure of brain tissue strain that occurs during an impact and is calculated using six acceleration-time curves and a finite element (FE) brain model. A finite element brain model includes the essential anatomical structures of a human head including skin, scalp, cerebrospinal fluid, brain medulla, spinal cord, cervical vertebrae, and discs. The FE brain model is used to determine the resulting brain tissue strain outcome from a head impact. MPS indicates the highest magnitude of strain from an impact to the brain tissue. Maximum principal strain (MPS) has been employed to measure injury risk (Clark et al, 2016). Injury reconstructions of head impacts in sport have been compared to validated FE brain models (Kleiven, 2007; Zhang et al, 2004; Clark et al, 2020) and determined that the resultant brain tissue strain (MPS) obtained from impacts can be correlated to the likelihood of sustaining a concussive injury (Kleiven, 2007; Zhang et al, 2004; Clark et al, 2020). It was reported that MPS values of 0.18 and above result in an estimated 50% chance for concussive injury (Zhang et al, 2004; Kleiven, 2007; Bain and Meaney, 2000; Post et al, 2019; Clark et al, 2020). Table 1 shows the concussion risk category and the MPS magnitude reported for each category (Bain & Meaney, 2000; Clark et al, 2020; Karton & Hoshizaki, 2018; Kleiven, 2007; Zhang et al., 2008).

Table 1:

MPS magnitude per concussion risk category.

Risk Category	MPS Magnitude
Very low	0-7.99%
Low	8-16.99%
Medium	17.0-25.99%
High	26.0-34.99%
Very high	35%+

Note. (Bain & Meaney, 2000; Clark et al, 2020; Karton & Hoshizaki, 2018; Kleiven, 2007; Zhang et al., 2008)

Impact Characteristics in Equestrian

Impact locations create different dynamic responses, and consequently, different brain tissue responses (Tiernan & Byrne, 2019; Elkin et al, 2019). Tiernan & Byrne (2019) and Elkin and colleagues (2019) used FE modeling and reported that lateral impacts result in significantly higher brain strains in the corpus callosum. Forero Rueda & Gilchrist (2010) reported impacts to the side of the head at a 45-degree angle resulted in higher brain tissue strain values when compared to impacts to the front and rear of the head. For frontal head impacts, Tiernan and Byrne (2019) reported the brainstem and the midbrain result in the highest brain strains for this impact location. For rear head impacts, inferior sections of the brain stem resulted in the highest brain strain (Tiernan & Byrne, 2019). For a rotational technology to be effective it must work in multiple directions to protect the brain regardless of impact locations.

Head impacts in equestrian events involve both high compliant (i.e., sand or turf) and low compliant surfaces (i.e., concrete). High compliant surfaces deform to absorb energy (Clark et al, 2020), resulting in lower magnitude accelerations (30.2-135.5g) and longer duration impacts when compared to less compliant surfaces (Clark, Connor et al, 2020; Clark, Hoshizaki, Annaidh, and Gilchrist, 2020). Hoshizaki, Post and colleagues (2017) reported long impact durations paired with low acceleration magnitudes can result in high levels of brain tissue strain. Brain tissue is viscoelastic in nature, making it sensitive to the duration of time acceleration is applied during an impact; this influences the resulting tissue strain (Post, Hoshizaki, Gilchrist et al, 2017). Clark, Connor, and colleagues (2020) concluded equestrian helmets should be tested on both high compliant surfaces and low compliant surfaces since they both carry risk for brain tissue structural damage. Equestrian helmets that are tested on real-world compliance impact surfaces may provide improved level of protection under real-world conditions.

Helmet Testing Standards

Helmets are primarily designed to mitigate high linear forces and protect against focal brain injuries (Hoshizaki & Brien, 2004). Helmets are tested for certification using a linear drop to a steel anvil. The incidence of focal brain injuries is low; however, the concussion incidence has not decreased, suggesting current testing methods do not sufficiently represent real-world conditions for concussive impacts. As a result, helmets are not designed to manage the loading conditions that reflect real-world accidents (Clark, Adanty et al, 2020). Equestrian helmets are tested using the protocols CEN1384:2016, ASTM F1163-15, and PAS015:2011. These tests involve a linear drop test of an unrestrained headform onto a steel anvil at the front, side, rear, and crown impact locations with an inbound velocity of 5.9m/s. These tests do not reflect real-world concussive accidents in equestrian events because: 1) falls are not typically a vertical drop as they have vertical and horizontal components, 2) many concussive equestrian events involve compliant surfaces such as turf or sand in equestrian (Clark, Adanty et al, 2020), and 3) impact velocities used in testing standards are too slow at 5.9m/s whereas the average impact velocity seen in equestrian accidents is 9m/s (Clark, Adanty et al, 2020). Consequently, working Group 11 of the European Committee for standardization has proposed CEN13087-11 to reduce the risk of concussions in equestrian. CEN13087-11 is an oblique (45 degree) impact test protocol to a steel anvil to three impact locations, front, side, and rear, at an impact velocity of 6m/s. Equestrian helmet testing standards are not reflective of real-world reconstructions of concussive events (Clark, Connor, et al, 2020; Clark, Adanty et al, 2020; Forero Rueda & Gilchrist, 2010). Drop tests are not an accurate representation of most falls in this sport, and the use of a rail guided launcher (RGL) may be useful to reconstruct real-world falls. Impacts to a highly compliant (i.e., turf or sand) surface are also not taken to consideration by this protocol. Highly compliant impacts have significantly different brain tissue strain outcomes than low compliant impacts in equestrian (Clark et al, 2020). Current helmet standards are primarily designed to protect the brain from linear impacts and do not adequately consider design changes that would offer more protection from oblique impacts. Helmet testing methods should consider real-world conditions when developing test methods to increase protection against both rotational and linear accelerations.

Clark and colleagues (2020) reconstructed 25 concussive and 25 non-concussive cases in equestrian sports. They reported impact velocities ranged from 6m/s to 12m/s, resulting in respective MPS values of 0.18 to 0.35 (Clark et al, 2021). Where an MPS value of 0.18 represents an approximately 50% chance of obtaining a concussion (Karton & Hoshizaki, 2018; Post et al, 2019). Clark and colleagues (2020) reviewed 50 real-world concussive and non-concussive equestrian accidents (Table 2) measuring the resultant impact velocities and corresponding trajectory angles. Clark and colleagues (2020) reported that the most common surface to be impacted in equestrian was a high compliant surface such as turf and/or sand. In comparison, an anvil compliance used for helmet testing is a low compliant surface such as steel. As seen in Table 2, the average impact velocity was 8.7m/s and the average trajectory angle was 27.45 degrees.

Table 2

Resultant impact velocities and trajectory angles from flat racing, jump racing, and cross-country events from Clark, Adanty and colleagues (2020) for equestrian.

Resultant Impact Velocity (m/s)	Trajectory Angle (degrees)
8.47	29
9.99	16
7.81	64
9.01	24
8.32	22
10.32	25
10.82	14
7.63	24
5.32	29
8.47	43

	3.93	21
	9.68	28
	13.48	23
	9.36	44
	7.72	21
	6.85	19
	9.74	21
	11.11	14
	15.24	14
	11.81	51
	6.58	22
	7.25	13
	6.4	23
	10.58	43
	13.32	17
	11.2	47
	11.56	10
	3.8	55
	7.7	26
	13.21	21
	6.28	43
	7.13	30
	6.81	29
	7.02	38
	9.15	21
	3.39	22
	9.75	20
	9.05	23
	16.24	18
	11.21	15
	7	20
	6.54	46
	11.59	13
	11.64	17
	9	33
	7.37	30
	7.51	34
	2.47	23
	6.74	47
	4.95	61
Average	**8.7**	**27.45**

Real-world equestrian reconstructions

Clark, Williams, and colleagues (2020) reported several brain injuries sustained in equestrian sports such as: hematoma, orbital fracture, fractured zygoma, fractured mandible, and concussion, where concussions presenting the highest percentage of cases (94.5%). The common injury events included fall, kick, collision, stomp, and crush, with falls representing the highest percentage of cases (78.5%). Another study by Connor, Clark et al (2019) reported head injuries as: 91% concussion, 4% skull fractures, 1% subdural hematoma, 1% cerebral edema, and 5% diffuse axonal injury.

The impact locations as reported by Clark, Williams, and colleagues (2020) were front boss (25%), side (26%), front (21%), and rear boss (16%). No impacts occurred to the crown of the helmet. Connor, Clark, Stewart, and colleagues (2021) reported that impacts to the crown were least common (17%). Connor, Clark, Stewart, and colleagues (2021) and Clark, Williams, and colleagues (2020) reported that the most common impact locations for an equestrian was front (21-25%), side (19-26%), and rear boss (16%).

Helmets are designed to deform to absorb energy from the head impact at a designated energy range. Clark, Connor et al (2020) reported after inspecting the equestrian helmets after a concussive fall accident, 53% of the helmets did show in any structural damage. Helmets that did not deform failed to absorb energy from the impact, which suggests the material may have been too stiff to offer protection (Becker,1998; Avalle et al, 2001). This highlights the need for helmets to be investigated with real-world concussive impact conditions to improve the protective capacity of helmets under real-world loading conditions.

Clark, Connor, Post, and colleagues (2020) reported that GoingRating (GR) is used to measure whether a racetrack is suitable for racing without risk of injury to the horse. The GR measures the compliance of the impact surface used in equestrian events. Anvils with different compliant surfaces resulted in significantly different head impact responses (Clark et al, 2020). Real-world concussive conditions are required to investigate the protective capacity of the helmets and the loading characteristics associated with these impacts.

Proposed new helmet testing protocol

In 2020, Clark, Hoshizaki, Annaidh, and Gilchrist proposed a novel test protocol using the monorail drop test. The impact anvil was set at 45 degrees to reflect current equestrian helmet standard CEN13087-11. Clark, Connor, Post, and colleagues (2020) used a range of vinyl nitrile (VN) foams to develop surrogate turf anvils. Three anvils were developed with varying compliances to reflect similar linear acceleration magnitudes and impact durations to represent racetrack compliance measure (Clark, Connor, Post et al, 2020). The surrogate anvils were developed by impacting a head on the front, side, and rear impact locations at 5.4m/s using a monorail drop rig. The anvils were impacted three times per impact conditions. The results of the developed surrogate anvils were specified to the targeted range for linear acceleration, rotational acceleration, and impact duration (Clark, Connor, Post et al, 2020). Clark, Connor, and colleagues reported that 66mm of 602VN foam with synthetic grass represented a mid-range compliance for turf impacts (2020). Clark and colleagues (2020) reported that the anvil surrogate can be used for multiple impacts and may be suitable for certification standard testing. The impact durations on turf surfaces for rotational accelerations (*Figure 1*), and linear acceleration (*Figure 2*) are presented below. There is an increased risk of brain injury with long impact durations, and therefore an association with brain tissue strain. Therefore, these anvils were used to adopt real-world compliances in equestrian.

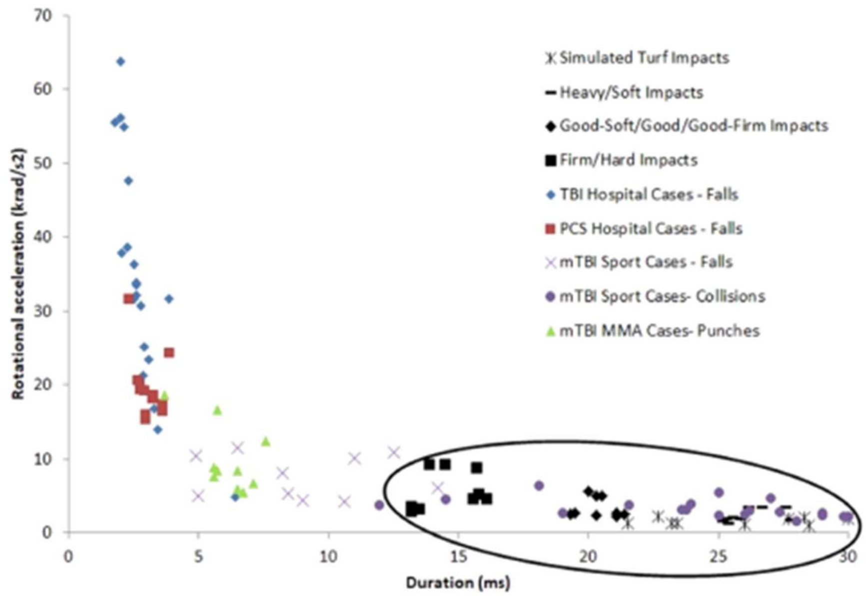

Figure 1. Rotational acceleration magnitude/duration relationships for impacts to turf and reconstructions of brain injury reported in the literature. Reprinted from "Could a Compliant Foam Anvil Characterize the Biofidelic Impact Response of Equestrian Helmets?", by M Clark, 2020, Journal of biomechanical engineering, 142(6), p8. Copyright [2022] by the name of Michio Clark. Reprinted with permission.

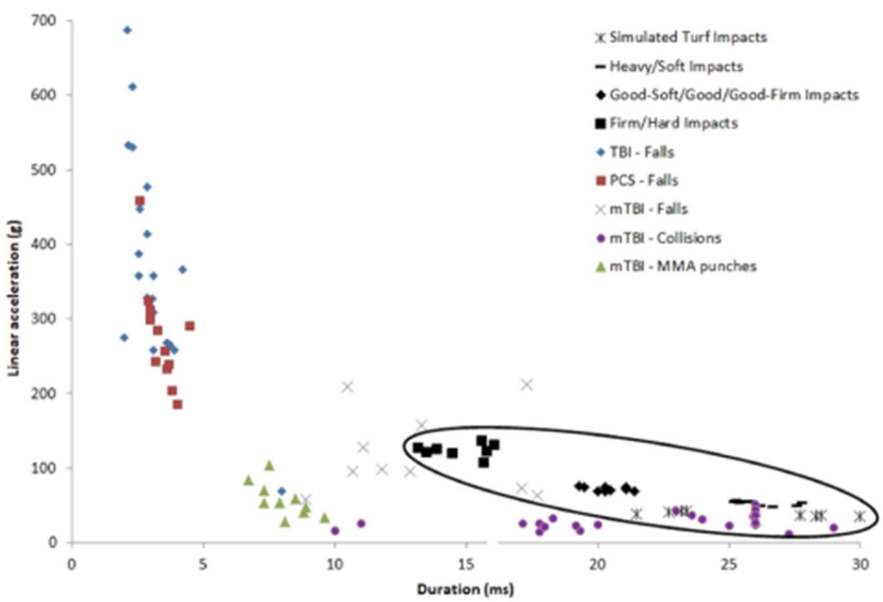

Figure 2. Linear acceleration magnitude/duration relationships for impacts to turf and reconstructions of brain injury reported in the literature. Reprinted from "Could a Compliant Foam Anvil Characterize the Biofidelic Impact Response of Equestrian Helmets?", by M Clark, 2020, Journal of biomechanical engineering, 142(6), p7. Copyright [2022] by the name of Michio Clark. Reprinted with permission.

Helmet Rotational Technology

Technology has been created and designed to be in helmets to mitigate rotational accelerations on head impacts and subsequently reduce the risk for concussions. Examples of rotational mitigation technologies include Fox Fluid Inside, Spin, Turbine by Leatt, Kali Protectives Low Density Layer (LDL), WaveCel, Koroyd, SHRED Rotational Energy System, and multi-directional impact protection system (MIPS). Many of these technologies were developed using a 45-degree impact to a steel anvil, how these technologies perform under sport specific conditions are not known.

The Fox Fluid Inside helmet is said to mimic cerebrospinal fluid to manage rotational and linear energy that causes concussions. Upon impact, the foam compresses and the low shear fluid is dispersed throughout the pod to manage linear and rotational acceleration. SPIN helmets are made of individual silicon gel pads placed strategically around the inside of the helmet designed to manage rotational accelerations. On head impact, the SPIN pads shear or slide to disperse the rotational force. Turbine by Leatt uses small rubber like turbine shaped circles placed strategically around the interior of the helmet. On head impacts, the material detects stress and hardens to absorb the impact energy before returning to original state. Kali Protectives LDL uses a similar silicon gel to SPIN, where they are placed around the inside of the helmet. Kali Protectives LDL gel pockets harden as they compress and move side to side to displace rotational energy upon head impacts. WaveCel is a helmet technology where the material incorporates a collapsible cell structure that acts in three distinct phases: on head impact, the WaveCel will flex to reduce frictional forces, crumble to slow linear forces, then glide to mitigate rotational forces. Koroyd uses a tubular material that is lightweight and comprised of straw-like co-polymer dual core tubes that are stuck together. When Koroyd is impacted, the tubes collapse in a consistent and predictable manner to disperse the force among other tubes to slow linear acceleration. Koroyd reduces rotational motion as a by-product of the way the material collapses. SHRED Rotational Energy System uses individually placed dots to reduce rotational acceleration; these small dots are comprised of little discs that move independently and multi-directionally to reduce rotational acceleration on head impacts.

Multi-directional impact protection system (MIPS) uses a low friction layer to allow the head to move 10-15mm relative to the helmet in all directions (Aare & Halldin, 2003; *MIPS® Safety System functionality,* 2021). The outer shell is made of a plastic or composite material. Under the shell are the layers of expanded polystyrene or expanded polypropylene foam. The foam functions to absorb energy from an impact. There is a sliding facilitator that allows for sliding between the low friction energy liner and the outer shell (*MIPS® Safety System functionality,* 2021). The movement of the head is intended to reduce rotational motion of the brain to reduce the amount of brain tissue strain (Aare & Halldin, 2003). The current study included MIPS rotational technology in equestrian helmets.

Helmets with MIPS technology are tested using a drop test onto an oblique steel anvil covered in 80-grit grinding paper at an impact velocity of 6.2m/s (*The test lab,* 2021*; Test angles and impact locations,* 2021). An angle of 45 degrees is used in MIPS testing because it represents the most common impact angle in bike accidents, and the helmets are under the most stress at 45 degrees (*The test lab,* 2021*; Test angles and impact locations,* 2021). A Hybrid III headform made of polyurethane and metal inserts instrumented with a 9-accelerometer array was used for all tests. The helmets were tested by impacting the front, side, and pitch impact locations. These impact locations were chosen because most head impacts occur within those regions (*The test lab,* 2021).

The purpose of these rotational technologies is to reduce the magnitude of head rotational acceleration sustained during impact, reducing the level of strain experienced by of brain tissue. Research has shown that helmets with the use of rotational managing technologies reduces head rotational acceleration which are associated with the risk of concussion (Adayazid et al, 2011; Bottland et al, 2020; DiGiacomo& Bottlang, 2021). Literature reported that rotational managing technologies can be very effective under different test methods and varying rotational technology construction (Adayazid et al, 2011; Bottland et al, 2020; DiGiacomo& Bottlang, 2021). Results from Adayazid and colleagues (2011), Bottland and colleagues (2020), and DiGiacomo and Bottlang (2021) show that characteristics of real-world accidents may not be accurately represented in current testing methods of equestrian helmets. Current test methods for these helmets do not incorporate the average impact characteristics, such as impacts to a high compliant surface, impact velocities higher than 6m/s, or impact trajectories at 27 degrees which were reported by Clark and colleagues (2020).

A study by Adayazid and colleagues in 2021 assessed a series of bicycle helmets with various rotational managing technologies to conventional helmets to see whether helmets with rotational managing technology provided better brain protection than conventional helmets. Their study used a helmet test protocol with a helmeted headform dropped onto a 45-degree incline anvil at 6.3m/s to three impact locations (Adayazid et al, 2021). Adayazid and colleagues reported helmets with rotational managing technology did considerably better than conventional helmets with respect to reducing peak rotational acceleration, peak rotational velocity, and maximal strain in the corpus callosum and sulci (2021). Adayazid and colleagues' study reported certain helmets with rotational managing technology were better at reducing strain in impacts under certain conditions (2021). These results highlight the importance of sport specific testing, and an understanding of how sports specific concussions occur and the variables that describe them are needed to better guide helmet design and innovation to improve the helmet protection against concussive injuries (Hoshizaki et al, 2014).

Helmet testing using real-world concussion impact conditions is required to establish whether helmets with rotational technology significantly reduces rotational acceleration. Helmets tested using the current standards do not reflect real-world accidents in equestrian, nor does the existing vertical drop test onto a low compliant anvil reflect falls in equestrian. Therefore, how helmets with rotational technology perform on real-world equestrian concussive accidents is still unknown.

Research Question

Do equestrian helmets with rotational technology significantly reduce peak linear and rotational accelerations, peak rotational velocities, and MPS when compared to helmets without rotational technology in an equestrian specific helmet test protocol?

Methodology

Objective

The objective of this research was to investigate the efficacy of MIPS rotational technology in managing linear and rotational acceleration, rotational velocity, and MPS using an equestrian specific test protocol.

Variables

Independent Variables

1) Jockey helmet type (2)
    a. with rotational technology
    b. without rotational technology

2) impact anvils (2)
    a. low compliance
    b. high compliance
3) impact location (3)
    a. front
    b. side
    c. rear boss

Dependent Variables

1) peak resultant linear acceleration
2) peak resultant rotational velocity
3) peak resultant rotational acceleration
4) peak MPS

Hypothesis

Null

1) Helmets with and without rotational technology will not have significantly different peak MPS values when tested on an equestrian specific protocol.
2) Helmets with and without rotational technology will not have significantly different peak resultant rotational velocity values when tested on an equestrian specific protocol.
3) Helmets with and without rotational technology will not have significantly different peak resultant rotational acceleration values when tested on an equestrian specific protocol.
4) Helmets with and without rotational technology will not have significantly different peak resultant linear acceleration values when tested on an equestrian specific protocol.

Research design

A1,2,3= Location: front (1), side (2) and rear (3)

B1,2=Compliance: low (1), high (2)

C1,2=Helmet: rotational technology (1), no rotational technology (2)

Table 3.

Research design for methodology.

	B1	B2	
A1	A1B1C1 A1B1C2	A1B2C1 A1B2C2	C1 C2
A2	A2B1C1 A2B1C2	A2B2C1 A2B2C2	C1 C2
A3	A3B1C1 A3B1C2	A3B2C1 A3B2C2	C1 C2

Statistical Analysis

To examine an overall difference between the helmet with and without rotational technology, four t-tests were performed with the data collapsed across anvil compliance (low and high), and impact location (front, side, and rear boss). The data was then separated by anvil type (low and high compliance) and four 2x3 fully crossed analysis of variance (ANOVA) with the factors helmet type (2 levels: with and without rotational technology), and impact location (3 levels: front, side, and rear boss) were performed. Eight pairwise post hoc tests (4 for low compliance and 4 for high compliance) were used to identify significant differences between the impact locations for each level of the independent variables: peak resultant linear acceleration, peak resultant rotational acceleration, peak resultant rotational velocity, and peak MPS.

Significance was set at α=0.05. The statistical software package used in this study was IBM SPSS Statistics 19 for Windows.

Test Procedure

A testing protocol to reflect real-world equestrian conditions observed by Clark and colleagues (2020) was used for the current study. Head impacts were performed using a 50^{th} percentile Hybrid III head form (4.54 ± 0.01kg) to represent a human head. The helmets had a circumference of 60cm to properly fit the 50^{th} percentile male headform. The headform SLICE NANO free motion data acquisition system was equipped with a triaxial accelerometer with three angular rate sensors mounted at the head center of gravity (DTS 6DX0399 PRO MODEL, Diversified Technical Systems, Seal Beach, CA). Accelerometer signals were passed through a TDAS Pro Lab system [DTS, Calabasas CA] before being processed by TDAS software. The data obtained was linear acceleration (g), rotational acceleration (rad/s^2), and rotational velocity (rad/s) time histories about the X, Y, and Z axis of the headform. The measures were sampled at 20 kHz and filtered with a 300 Hz filter (CFC 180). The Hybrid III headform was fitted with commercially available Charles Owen jockey equestrian helmets. The helmets tested was the MS1 Pro with MIPS technology and without MIPS technology.

The rail guided launcher (RGL) (*Figure 5*) was used to represent falling in a projective parabola compared to vertical drops (Clark, Adanty et al, 2020) to test helmet performance capacity in conditions that reflect real-world accident conditions. The RGL was angled at 26.5 degrees to the horizontal to represent the average trajectory angle reported by Clark and colleagues (2020) in their reconstructions of real-world equestrian accidents of 27.45 degrees. Two compliant anvils were used: a low compliance concrete anvil (*Figure 4a, b*) and a high compliant foam surface (*Figure 3*) with synthetic grass to compare the helmets with and without rotational technology to helmets on a real-world testing protocol. The high compliant anvil consisted of 66mm of 602 vinyl nitrile (VN) foam secured to the concrete floor by double sided carpet tape (*Figure* 3). The 602 VN foam was chosen to represent the mid-range compliance developed by Clark, Connor, Post, and colleagues in 2019 and most closely represented sand impacts (Clark, Adanty, Post et al, 2020). The foam was covered with synthetic grass (SYNGrass UltraLush ™ 40) to replicate compliant turf surfaces (Clark, Adanty, Post, et al, 2020). The high compliant anvil was reported to produce repeatable results over multiple impacts and would be suitable for standardized helmet testing to represent the level of a high compliance impact associated with a racetrack GR (Clark, Adanty, Post, et al, 2020).

Figure 3. The high compliant anvil made of 66mm of 602 vinyl nitrile foam with synthetic foam on top.

Figure 4a. The low compliant anvil that consisted of a bare concrete surface at the end of the rail guided launcher.

Figure 4b. The low compliant anvil superior view.

Figure 5. The rail guided launcher (RGL) used for all impacts.

The headform was outfitted with two types of equestrian helmets (with and without MIPS) and launched onto each anvil at three different locations: front (*Figure 3(F2),4*), side (*Figure 3(L5),5*), and rear boss (*Figure 3(R6),6*). A high-speed camera was used to capture each impact event to measure velocity and to ensure a consistent impact location.

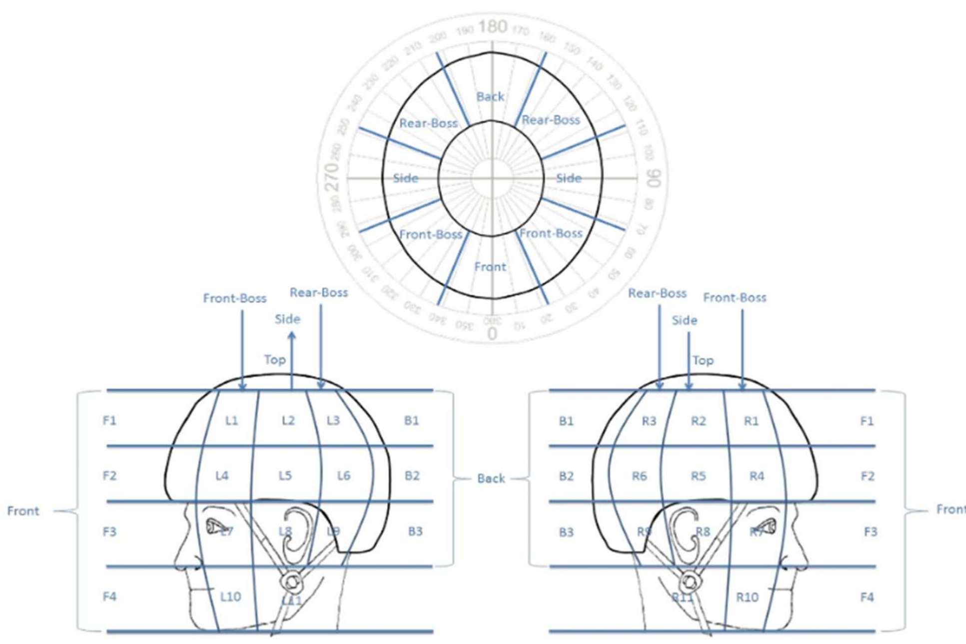

Figure 6. Left and right view of a helmeted head for the impact locations.

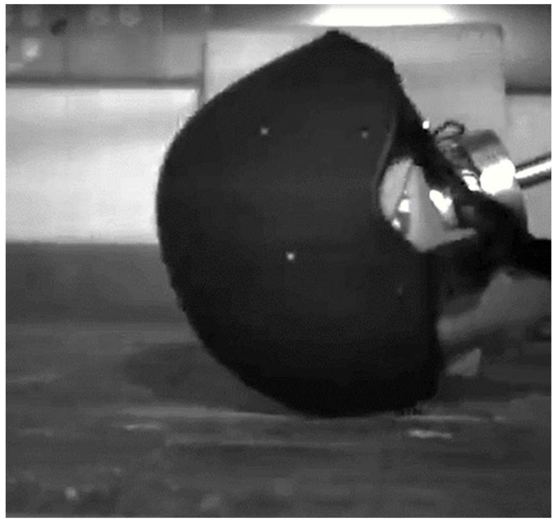

Figure 7. High speed video screenshot to the front impact location.

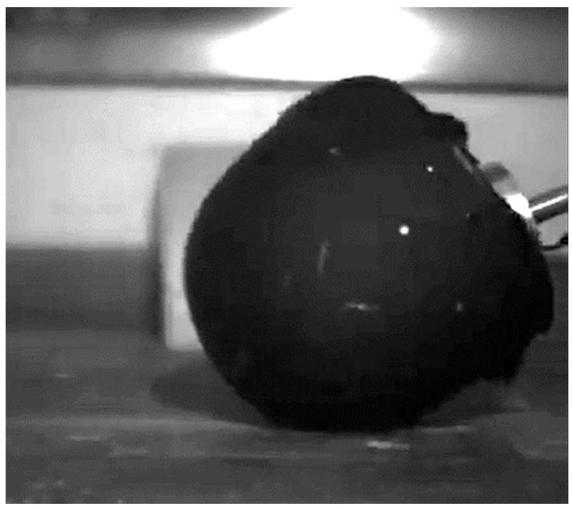

Figure 8. High speed video screenshot to the side impact location.

Figure 9. High speed video screenshot to the rear boss impact location.

Each helmet was impacted once per location (front, side, and rear boss), for a total of 3 impacts per helmet. Only three impacts were allotted because after repeated impacts equestrian helmets have reduced ability to attenuate energy and should be replaced (Mattacola, 2019). Each location was repeated 3 times and 6 helmets were used per anvil condition (i.e., three helmets with rotational technology, and three helmets without rotational technology; for both low and high compliance, for a total of 12 helmets). Each helmet was launched with an impact velocity of 9.0 ± 0.25 m/s. The impact velocity was 9m/s and was chosen as it represents the average impact velocity of reconstructed equestrian head impacts, the 50% probability of sustaining as determined by Clark et al (2020), and represents a 0.24 MPS value and within the range for a 50% probability of sustaining a concussion (Clark, Adanty, Post, Hoshizaki, and Clissold al, 2020). The outlined velocity falls within the brain tissue strain range of brain tissue structural damage (>0.18 MPS) (Zhang et al, 2004; Kleiven, 2007; Bain and Meaney, 2000; Post et al, 2019).

Helmet Construction

The helmet with rotational technology was the MS1 Pro with MIPS by Charles Owen, and the helmet without rotational technology was the MS1 Pro by Charles Owen with the MIPS technology removed. The MS1 Pro outer shell was made of Acrylonitrile Butadiene Styrene (ABS) and the inner foam was Expanded Polystyrene (EPS). The outer shell was 2 mm thick, and the thickness of EPS varied throughout the helmet. The EPS foam was 28mm thick at the front impact location, 36mm at the side location, and 30mm at the rear boss impact location. The MIPS rotational technology was installed on the inner surface of the EPS layer. A removable headband was secured to the rim of the helmet via Velcro. This helmet has been certified using the following standards: ASTM F1163-15, Kitemarks to VG1 01-040 2014-12 and PAS015:2011, and a CE mark to VG1 01-040 2014-12.

Figure 10. MS1 Pro Charles Owen equestrian helmet with MIPS technology and headband removed.

Figure 11. MS1 Pro Charles Owen equestrian helmet without MIPS technology and headband removed.

Finite Element Modeling

The University College Dublin Brain Trauma Model version 2.0 (UCDBTM V2.0) was used to complete finite element (FE) analysis for each impact. The FE model was used to quantify brain tissue strain, specifically maximum principal strain. The UCDBTM V2.0 model was based on the analysis of linear and rotational acceleration curves produced from an impact (Horgan & Gilchrist, 2003; King et al., 2003; Ommaya et al., 1966; Post et al., 2017). Version 2.0 was developed by Trotta and colleagues (2020) to incorporate the addition of new mechanical properties and sliding properties of the scalp and modified mechanical properties of the brain.

The UCDBTM V2.0 has been validated against two sets of cadaver data. The data by Loyd et al (2011, 2014) was used to validate the scalp model, and the data reported by Hardy et al (2001, 2007) was used to validate displacement in the brain. The UCDBTM V2.0 was compared against the UCDBTM V1.0 on concussion thresholds by Clark et al (2018) with regards to equestrian data, suggesting that the UCDBTM V2 is suitable for the use of equestrian research.

The mechanical properties of the scalp depend on strain rate and the direction of the impact at quasi-static speeds. At high strain rates the scalp behaves isotopically, and a hyper-elastic Ogden model was used to describe the scalp (Trotta & Annaidh, 2019). Further improvements have been made by MacManus and colleagues (2017) from version one regarding the constitutive parameters that describe the mechanical properties of the cerebellum, the brainstem, and grey matter from mechanical properties of 20–25-week-old rats. It was accepted that the mechanical properties of 20–25-week-old rats corresponded to adult human brain tissue (MacManus et al, 2017; Sengupta, 2013). FE head models include linear descriptions of dura mater, where current work has reported that the dura mater is a hyper-elastic material that the stiffness has been overestimated by 45% (Ho et al, 2017). Work by Van Noort and colleagues (1981) reported that the mechanical properties of the dura mater depend on age of the subject where stiffness reduces with age. The same mechanical properties were used to model the falx and the tentorium. To replicate the physiological boundary of the scalp more closely, a surface-to-surface contact that uses a penalty contact method that does not allow for separation after contact was added to the FE model to allow for the scalp to slide over the skull with no penetration between the two surfaces (Trotta et al, 2020). The scalp-skull friction coefficient was 0.06 and was made to replicate a realistic simulation of the behaviour of the scalp (Trotta et, al, 2020; Trotta et al, 2018). The scalp or helmet interaction with the ground or impact surface was set to 0.5 in line with other FE head models (Fahlstedt, Halldin, and Kleiven, 2016). The coefficient of friction between the scalp and the absorbing helmet liner was set to 0.3 based on work by Trotta and colleagues (2018). All other interactions between the pia mater and cerebrospinal fluid had a coefficient of friction of 0.2 (Horgan & Gilchrist, 2003). There were several changes to improve the model performance, reduction of element distortion, and to obtain more accurate results. Hypermesh was refined from 28,286 to 184,261 elements to improve the Jacobian value of the mesh elements (Trotta et al, 2020; Trotta et al, 2018). The Jacobian ratio was a measure of the deviation of an element from an ideally shaped element. To reduce the distortion of the elements and to maintain element quality, Arbitrary Lagrangian-Eulerian adaptive meshing was used for the scalp (Trotta et al, 2020). Accelerometer elements were added to the center of gravity to report linear and rotational accelerations undergone by the head during an impact. The summary of mechanical properties for all other tissues of the UCDBTM v2.0 are in Table 4.

Table 4.

Mechanical properties of the UCDBTM Version 2.0. Reprinted from “Biofidelic finite element modelling of brain trauma: Importance of the scalp in simulating head impact” by Trotta, Clark, McGoldrick, Gilchrist, and Annaidh, 2020. International journal of mechanical sciences, 173, p2.

Region	Model	Density [kg/m3]	Poisson's ratio	Parameters
Scalp [27]	Hyperelastic (Ogden)	1133	~ 0.5	$\mu = 1.48$ MPa $\alpha = 8.1$
Cerebellum [28]	Visco-hyperelastic	1060	~ 0.5	$\mu = 2611$ Pa $g_1 = 0.515$, $t_1 = 0.020$ s $g_2 = 0.187$, $t_2 = 0.302$ s $g_{inf} = 0.298$
Grey matter [28]	Visco-hyperelastic	1060	~ 0.5	$\mu = 5715$ Pa $g_1 = 0.534$, $t_1 = 0.020$ s $g_2 = 0.207$, $t_2 = 0.304$ s $g_{inf} = 0.258$
Brainstem [28]	Visco-hyperelastic	1060	~ 0.5	$\mu = 4768$ Pa $g_1 = 0.63$, $t_1 = 0.0185$ s $g_2 = 0.175$, $t_2 = 0.290$ s $g_{inf} = 0.290$
Cortical bone [4]	Linear elastic	2000	0.22	E = 1500 MPa
Trabecular bone [4]	Linear elastic	1300	0.24	E = 1000 MPa
Pia [4]	Linear elastic	1130	0.45	E = 11.5 MPa
CSF [4]	Linear elastic	1000	~ 0.5	E = 0.15 MPa
Facial bone [4]	Linear elastic	2100	0.22	E = 5540 MPa
Ventricles [4]	Visco-hyperelastic	1040	~ 0.5	C10 = 3653.5 Pa C01 = 4059.44 Pa $g_1 = 0.527$, $t_1 = 0.008$ s $g_2 = 0.303$, $t_2 = 0.145$ s
White matter [4]	viscoelastic	1060	~ 0.5	E = 37,500 Pa g_1 0.8, t_1 0.0125 s
Dura, falx and tentorium [31]	hyperelastic	1130	~ 0.5	$\mu = 3.602$ MPa $\alpha = 13.73$

Exclusion Criteria

All trials where the headform rotated before impacting were discarded and excluded. All trials with a sensor error were excluded. All trials where the impact velocity was outside of 9.0 ± 0.25 m/s were excluded.

Results

Helmet Comparison

The data was collapsed across anvil compliance and impact location and separated by helmets with and without rotational technology. A t-test was used to examine if there was a difference between the two means of helmets with and without rotational technology. The t-test revealed no significant differences between the means of the two helmets when the data was collapsed across anvil compliance and impact location for all the dependent variables (*Figure 12*).

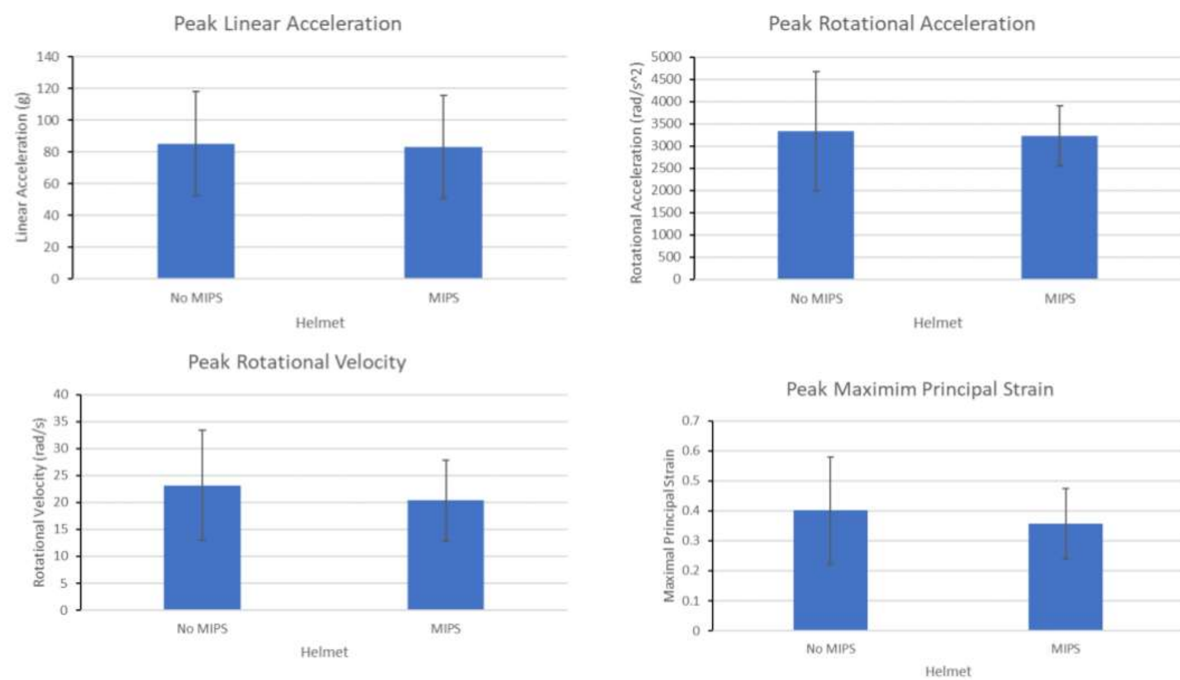

Figure 12. Average peak linear acceleration, peak rotational acceleration, peak rotational velocity, and MPS for helmets with and without rotational technology.

<u>Low Compliance Anvil</u>

The data was further explored by examining the low compliance anvil using a 2x3 ANOVA. The 2x3 ANOVA revealed significant differences, and a pairwise post hoc test was used to determine the significance between the impact locations. The pairwise post hoc revealed significant differences between the helmets with and without rotational technology for peak resultant rotational acceleration (p=0.005), peak resultant rotational velocity (p=0.004), and peak maximum principal strain (p=0.002) at the rear boss impact location, where the helmet with rotational technology had a lower peak rotational acceleration than the helmet without rotational technology (*Figure 13*). There were significant differences between helmets with and without rotational technology for peak maximum principal strain at the front impact location ($p<0.001$) where the helmet with rotational technology had significantly higher MPS values than the helmet without rotational technology.

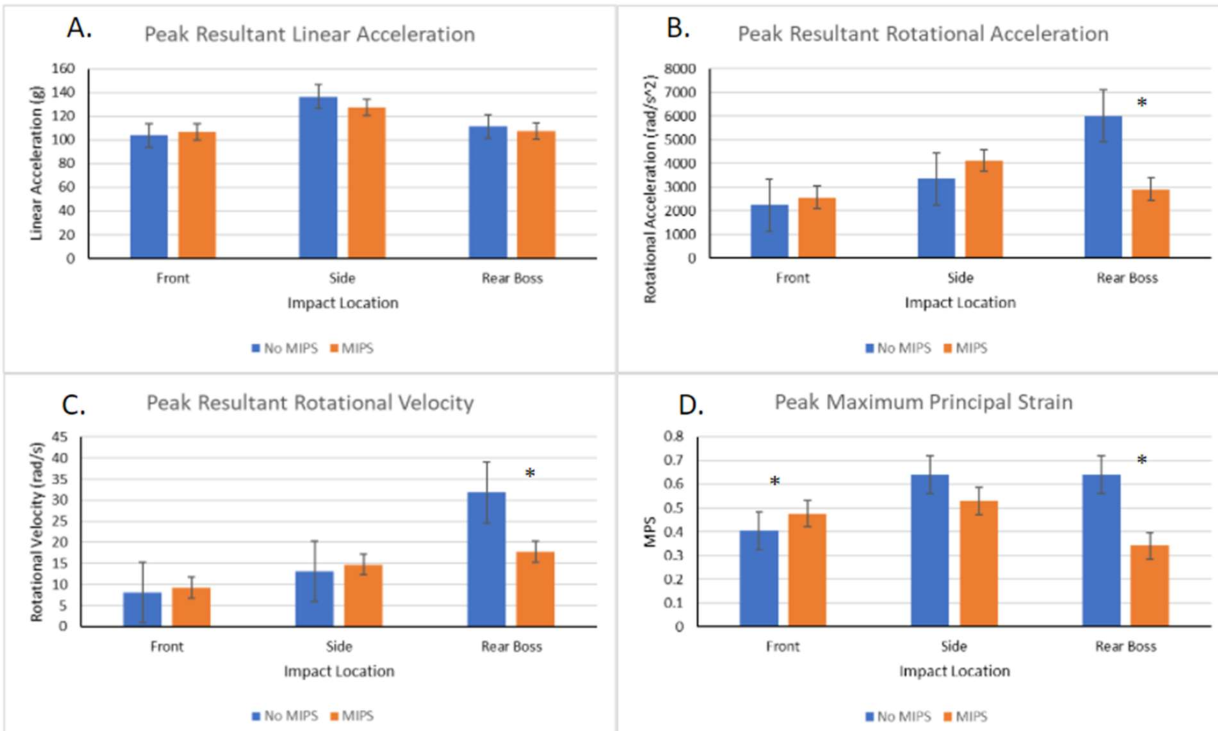

Figure 13. Low compliance impacts for helmets with and without rotational technology results for peak linear acceleration, peak rotational acceleration, peak rotational velocity and peak maximal principal strain for front, side, and rear boss impact sites. * Represents significant effects ($p<0.05$).

<u>High Compliance Anvil</u>

The 2x3 ANOVA for the high compliant conditions revealed significant differences and a four pairwise post hoc test were used to determine significance for impact site. The Tukey post hoc t-tests revealed significant peak resultant linear acceleration differences between the helmets with and without rotational technology at the front ($p=0.004$) and rear boss ($p=0.004$) impact locations, where the helmet with rotational technology had lower peak linear acceleration values (*Figure 14A*). There were no significant differences found at the side impact location for peak linear acceleration. There were no significant differences found for peak resultant rotational acceleration between the helmets with and without rotational technology (*p*=0.165). (*Figure 14B*). There were significant differences found for peak resultant rotational velocity between the helmets with and without rotational technology at the rear boss impact location (*p*=0.012), where the helmet with rotational technology had significantly lower peak rotational velocity values than the helmet without rotational technology (*Figure 14C*). The other impact locations were not significantly different with respect to peak resultant rotational velocity. There were significant differences found for peak MPS between the helmets with and without rotational technology for the side impact location (*p*=0.001), where the helmet with rotational technology had significantly higher MPS values than the helmet without rotational technology (*Figure 14D*). The other impact locations were not significantly different between the helmets with respect to MPS values.

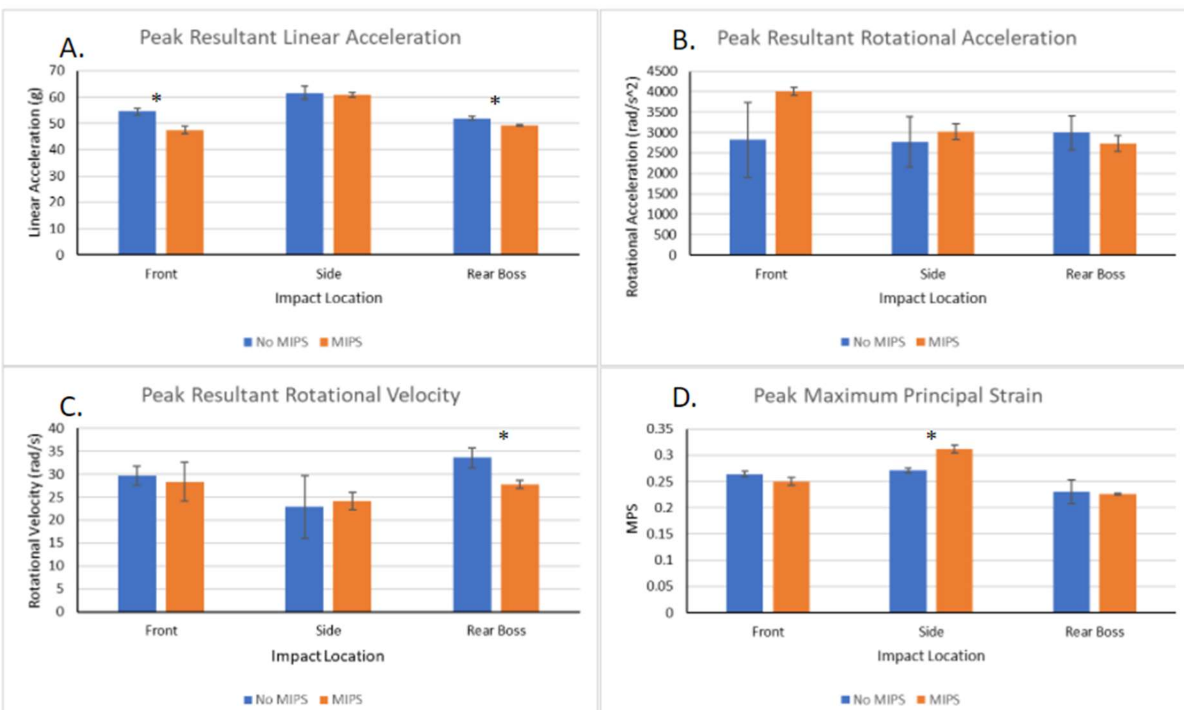

Figure 14. High compliance impacts for helmets with and without rotational technology results for peak linear acceleration, peak rotational acceleration, peak rotational velocity and peak maximal principal strain for front, side, and rear boss impact sites. * Represents significant effects ($p<0.05$).

Discussion

This research compared helmets with and without rotational technology, specifically MIPS technology, using a test protocol representing real-world head impact conditions in equestrian events that led to concussive injuries. The purpose of this research was to evaluate whether rotational technology in equestrian helmets reduce brain tissue strain (peak MPS) and dynamic head responses (peak resultant linear acceleration, peak resultant rotational acceleration, and peak rotational velocity) when compared to helmets without rotational technology under real-world test conditions. The two helmets used in this research were the MS1 Pro with MIPS and the MS1 Pro without MIPS. When the conditions were collapsed across anvil compliance and impact locations, the helmet with and without rotational technology were not significantly different from one another with respect to the dynamic head response and brain tissue strain. Further analysis was done to examine the differences in performances between the two helmets on a low and high compliant anvil. The performance of each helmet type varied by impact location.

MPS values for the low and high compliant anvils were above the estimated 50% risk for concussion (Zhang et al, 2004; Kleiven, 2007; Bain and Meaney, 2000; Post et al, 2019; Clark et al, 2020). MIPS helmet technology was most effective for the rear boss impact location when compared to other impact locations for both anvils. The high compliant anvil resulted in lower MPS values when compared to the low compliant anvil supporting the importance of anvil compliance when assessing helmet performance.

The suspension on the MIPS rotational technology broke on all impacts excluding one trial, where at least one of the four connectors were broken. The suspension either became dislodged from the liner of the helmet (*Figure 15*), or the suspension broke midway between the insertion on the helmet and the liner (*Figure 16*). *Figure 17* shows what the suspension looked like when it was fully intact. The liner of the rotational technology appeared to be unaffected, however the suspension that holds the liner to the shell of the helmet became dislodged and torn. This in part may have been the result of the high impact velocity (9m/s) used in this research. This may have decreased the effectiveness of the MIPS technology to manage rotational accelerations and brain tissue strain. When the rotational technology was no longer secured to the helmet, it changed the interaction of the helmet and the liner and may have made the head more susceptible to an increased dynamic head response and brain tissue strain. The rotational technology could be more robust, so the connectors do not disengage at high energies. It is likely the rotational technology was beyond its functional range under these conditions. The helmets are tested at lower energy impacts and may provide effective protection under those conditions; however, the material may never have been examined under real-world high-energy conditions. The materials would be unknown how it relates to the fracture index and performance. A more secure technology designed to withstand energy levels that best reflect real-world conditions may improve rotational technology performance (McIntosh et al, 2011). These results support the use of real-world, sport specific test conditions when developing and evaluating sport helmets.

Figure 15. The suspension of the rotational technology when it dislodged from the liner.

Figure 16. The suspension of the rotational technology broken between the liner and the shell of the helmet.

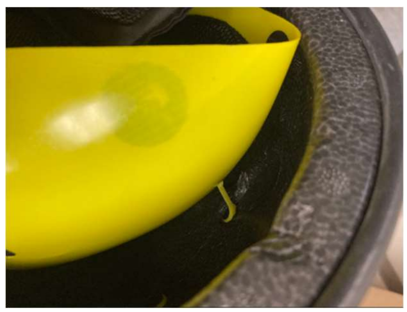

Figure 17. The suspension of the rotational technology connected between the liner and the shell of the helmet.

This research provided information and an understanding on how helmets with rotational technology perform on a real-world impact protocol in equestrian.

Helmet Comparison

The performance of the two helmets were not significantly different from one another when the data was collapsed across anvil compliance and impact location. Helmets with rotational technology did not perform consistently better for both anvil compliances and all impact locations. Further investigation examined how specific characteristics of the impact event influenced performance differences between the two helmet types.

Linear Acceleration

Commercial helmets are designed and tested to a certification standard with pass/fail criteria based on linear acceleration, typically below a 250g threshold. These helmet standards use a drop test to a low compliant anvil. Helmets with rotational technology should include a testing methodology specific to rotational acceleration. In this study there were no significant differences between the helmets with and without rotational technology for peak resultant linear acceleration for the low compliant anvil. This is consistent with previous studies involving rotational technology as all helmets are designed and tested under similar conditions to manage linear accelerations (Aare & Halldin, 2003; *MIPS® Safety System functionality, 2021*).

Differences were observed in peak resultant linear acceleration for impacts to the high compliant anvil where helmets with rotational technology significantly reduced peak resultant linear acceleration at the front and rear boss impact locations compared to helmets without rotational technology. The high compliant anvil was designed to represent the racetrack compliance in real-world impact conditions in equestrian (Clark, Connor, et al, 2020; Clark, Hoshizaki, Annaidh, and Gilchrist, 2020).

Rotational Acceleration and Rotational Velocity

Significant differences were found with respect to peak resultant rotational acceleration and peak resultant rotational velocity at the rear boss location for the low compliant anvil. The helmet with rotational technology had lower rotational acceleration and rotational velocity values when compared to the helmet without rotational technology. The rotational technology was the most effective at decreasing rotational acceleration at the rear boss impact location where the peak resultant rotational acceleration and rotational velocities were reduced by 3076.56 rad/s^2 and 13.97 rad/s respectfully. The rotational acceleration was reduced by more than 50% by helmets with rotational technology. The results at the rear boss impact location indicate that rotational technology effectively reduced rotational components to the brain. However, the other impact locations did not reflect similar results. The MIPS rotational technology was inconsistent in its ability to mitigate rotational acceleration upon different impact locations highlighting the importance of testing real-world protocols. Protocols that include real-world conditions from specific sporting events can improve the ability for the rotational technology to effectively manage rotational accelerations.

Maximum Principal Strain (MPS)

Helmets with rotational technology reduced MPS from 0.639 to 0.342 for the rear boss impact location for the low compliance anvil. These results are consistent with decreases observed in peak resultant rotational acceleration and velocity under similar impact conditions. The front and side impacts the helmet with rotational technology resulted in significantly higher peak MPS when compared to the helmet without rotational technology. The impact characteristics, geometry of the head, helmet and rotational technology influence the resulting dynamic response and brain tissue strain. These results reflect the complex relationship between impact conditions and the resulting head dynamics, emphasizing the importance to consider unique protective and performance characteristics specific to the impact event.

These results demonstrate that the rotational technology can be effective at reducing the magnitude of brain tissue strain, though consideration must be given to the magnitude of real-world impacts to effectively manage rotational accelerations. The low compliant anvil results indicate that the MPS values represent a higher risk of concussion compared to a high compliant anvil. The risk for concussion increases as impact duration increases, which is associated with high compliant impacts. The inconsistent results and physical failure of the rotational technology supports the importance of using real-world test methods to guide rotational technology designs for equestrian helmets.

Implications for Rotational Technology Testing

Rotational technologies in helmets are intended to reduce the magnitude of head rotational acceleration and rotational velocities sustained during impacts, subsequently reducing the magnitude of brain tissue strains. Helmets including rotational managing technologies reduce head rotational acceleration associated with the risk of concussions (Adayazid et al, 2011; Bottland et al, 2020; DiGiacomo & Bottlang, 2021). Research shows that rotational managing technologies can be effective under specific test methods (Adayazid et al, 2011; Bottland et al, 2020; DiGiacomo & Bottlang, 2021), however, the results from this study show that impact characteristics of real-world accidents may not be fully represented in current testing methods used for equestrian helmets.

Multiple factors need to be considered to effectively manage head trauma and concussions in sport such as, impact velocity, trajectory angle, and impact compliance (Benson et al, 2013). Several studies have reported on the importance of sport specific impact testing for effective head protection (McIntosh et al, 2011; Benson et al, 2013). The present study reported inconsistent performance depending on anvil compliance and impact location. Rotational technology performed most effectively on the high compliant anvil for the rear boss impact location (*Figure 8,9*). Upon impact the MIPS technology's suspension to the helmets broke, resulting in inconsistent results which may be a result of the real-world conditions seen in this study. The test protocol used in this study employed a compliant impact surface, high impact velocity (9.0m/s) and a low impact trajectory (27 degrees) as reported by Clark and colleagues (2020). Sport specific helmets tested under impact characteristics representing real-world conditions and performance ranges can be used to target protection against specific injuries including concussions (McIntosh et al, 2011).

The findings in this study support the importance of testing equestrian helmets including rotational technologies using real-world parameters. Helmet technologies designed to manage rotational accelerations during head impacts are important, and this study provides useful information to show that future protocols should consider an angle trajectory of 26.5 degrees instead of 45 degrees.

Future Research

Future studies should include further investigations involving appropriate real-world impact velocities, compliance, and angles to capture the range of loading conditions the sport helmets sustain during accidents.

Conclusion

The dependent variables in this study included peak resultant linear acceleration, peak resultant rotational acceleration, peak resultant rotational velocity, and maximum principal strain. The purpose of this study was to investigate if equestrian helmet with rotational technology would reduce the dependent variables on an equestrian specific real-world test protocol compared to a helmet without rotational technology. The results indicate there were no significant differences between the helmets under these conditions when the data was collapsed across independent variables. The helmet with rotational technology reduced rotational accelerations and/or rotational velocity at the rear boss impact location. The results show that the helmet with rotational technology works well under certain conditions, however, it is not a universal improvement. This study offers insights to guide developing a protocol to test the ability of equestrian helmet to manage rotational acceleration by considering 26.5 degrees an anvil trajectory.

References

Aare M, Halldin P. (2003). A new laboratory rig for evaluating helmets subject to oblique impacts. *Traffic Injury Prevention.*4(3):240–248.

Abayazid, F., Ding, K., Zimmerman, K., Stigson, H., & Ghajari, M. (2021). A new assessment of bicycle helmets: the brain injury mitigation effects of new technologies in oblique impacts. Annals of biomedical engineering, 49(10), 2716-2733.

Avalle, M., Belingardi, G., & Montanini, R. (2001). Characterization of polymeric structural foams under compressive impact loading by means of energy-absorption diagram. *International journal of impact engineering*, *25*(5), 455-472.

Bain, A., Meaney, D. 2000. Tissue-level thresholds for axonal damage in a experimental model of central nervous system white matter injury. J Biomech Eng. 122:615-622.

Ball, C. G., Ball, J. E., Kirkpatrick, A. W., & Mulloy, R. H. (2007). Equestrian injuries: incidence, injury patterns, and risk factors for 10 years of major traumatic injuries. *The American Journal of Surgery*, *193*(5), 636-640.

Becker, E. B. (1998). Helmet development and standards. *Biomedical and Health Research, Frontiers in Head and Neck Trauma. California: IOS press*.

Benson, B. W., McIntosh, A. S., Maddocks, D., Herring, S. A., Raftery, M., & Dvořák, J. (2013). What are the most effective risk-reduction strategies in sport concussion?. British journal of sports medicine, 47(5), 321-326

Bottlang, M., Rouhier, A., Tsai, S., Gregoire, J., & Madey, S. M. (2020). Impact performance comparison of advanced bicycle helmets with dedicated rotation-damping systems. Annals of biomedical engineering, 48(1), 68-78.

Bradshaw DRS and Morfey CL (2001) Pressure and shear responses in brain injury models. In: 17th International Technical Conference on the Enhanced Safety of Vehicles, Amsterdam, The Netherlands.

Brain sensitivity and impacts. (2021, March 30). Mips. https://mipsprotection.com/science-technology/brain-sensitivity-and-impact/

Casson, L., Viano, D., Powell, J., & Pellman, E. (2010). Twelve years of National Football League concussion data. *Sports Health: A multidisciplinary Approach,*2(6): 471-83.

Clark, J. M., Adanty, K., Post, A., Hoshizaki, T. B., Clissold, J., McGoldrick, A., ... & Gilchrist, M. D. (2020). Proposed injury thresholds for concussion in equestrian sports. *Journal of science and medicine in sport*, *23*(3), 222-236.

Clark, J. M., Hoshizaki, T. B., Annaidh, A. N., & Gilchrist, M. D. (2020). Equestrian helmet standards: do they represent real-world accident conditions? *Annals of biomedical engineering*, *48*, 2247-2267

Clark, M., T. A. Connor, C. Williams, J. Clissold, A. McGoldrick, J. Hill, A. Ni Annaidh, and M. D. Gilchrist. (2020). Analysis of Helmet Damage and Associated Head Injuries Arising from Real-World Equestrian Fall Accidents. Journal of Testing and Evaluation 48, no. 3 :2185–2195. https://doi.org/10.1520/JTE20190513

Clark. M, Connor, T. A., Post, A., Blaine Hoshizaki, T., Ní Annaidh, A., & Gilchrist, M. D. (2020). Could a Compliant Foam Anvil Characterize the Biofidelic Impact Response of Equestrian Helmets?. *Journal of biomechanical engineering*, *142*(6).

Clark, M., Post, A., Hoshizaki, T et al. (2016). Protective capacity of ice hockey helmets against different impact events. Ann biomed Eng 44: 3693-3704.

Clark, J. M., Adanty, K., Post, A., Hoshizaki, T. B., Clissold, J., McGoldrick, A., ... & Gilchrist, M. D. (2018). Reconstruction of real world concussive and non-concussive accidents in equestrian sports. In *IRCOBI conference, Greece* (p. 2).

Connor, T. A., Clark, J. M., Jayamohan, J., Stewart, M., McGoldrick, A., Williams, C., ... & Gilchrist, M. D. (2019). Do equestrian helmets prevent concussion? A retrospective analysis of head injuries and helmet damage from real-world equestrian accidents. *Sports medicine-open*, *5*(1), 1-8.

Connor, T. A., Clark, J. M., Stewart, M., Annaidh, A. N., & Gilchrist, M. D. (2021). Post-accident evidence basis for new equestrian standards: Relationship between helmet liner residual crush and accident parameters. *Applications in Engineering Science*, *6*, 100044.

DiGiacomo, G., Tsai, S., & Bottlang, M. (2021). Impact performance comparison of advanced snow sport helmets with dedicated rotation-damping systems. *Annals of biomedical engineering*, *49*(10), 2805-2813.

Elkin, B. S., Gabler, L. F., Panzer, M. B., & Siegmund, G. P. (2019). Brain tissue strains vary with head impact location: A possible explanation for increased concussion risk in struck versus striking football players. *Clinical biomechanics*, *64*, 49-57.

EN 1384. Helmets for equestrian activities. British Standards Institution: European CE Standards, 2016.

Fahlstedt, M., Halldin, P., & Kleiven, S. (2016). The protective effect of a helmet in three bicycle accidents—A finite element study. *Accident Analysis & Prevention*, *91*, 135-143.

Forero Rueda, F., Cui, L., & Gilchrist, M. D. (2011). Finite element modelling of equestrian helmet impacts exposes the need to address rotational kinematics in future helmet designs. *Comput Methods Biomech Biomed Engin*, *14*(12), 1021-1031.

Forero Rueda, MA Forero, and M. D. Gilchrist. "Equestrian Helmet Design: A Computational and Head Impact Biomechanics Simulation Approach." *6th World Congress of Biomechanics (WCB 2010). August 1-6, 2010 Singapore*. Springer, Berlin, Heidelberg, 2010.

Gennarelli, T. A., Thibault, L. E., Tomei, G., Wiser, R., Graham, D. I., and Adams, J. (1987). “Directional dependence of axonal brain injury due to centroidal and non-centroidal acceleration,” in *Proceedings of the 31st Stapp Car Crash Conference* (Warrendale, PA: Society of Automotive Engineers).

Gennarelli, T. A. (1981), Mechanistic approach to the head injuries: Clinical and experimental studies of the important types of injury. head injury criteria - a concensus workshop, Technical report, U. S. Government Printing Office, Washington D. C.

Gennarelli TA, Thibault LE and Ommaya A (1972) Pathophysiological responses to rotational and translational accelerations of the head. In: 16th Stapp Car Crash Conference, Detroit, Ml, USA, SAE paper 720970.

Gennarelli TA, Thibault LE and Ommaya A (1971) Comparison of translational and rotational accelerations in experimental cerebral concussion. In: 15th Stapp Car Crash Conference, New York, USA.

Greenwald, R., Gwin, J., Chu, J., Crisco, J. 2008. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosurgery. 62:789-798.

Gurdjian ES (1975). Impact head injury: mechanistic, clinical and preventative correlations, Charles C Thomas.

Gurdjian ES, Lissner, HR, Latimer FR et al. (1958). Quantitative determination of acceleration and intracranial pressure in experimental head injury. Neurology 3: 417-423.

Gurdjian ES, Lissner HR, Patrick LM (1963). In: Concussion mechanism and pathology. Proceedings of the 7th Stapp Car Crash Conference. Pp 370-482.

Halldin, P. (2015). Proposal for a new test method measuring the head kinematics in angled helmeted impacts. Work on CEN TC158-WG11.

Halldin, P., Gilchrist, A., Mills, MJ. A new oblique impact test for motorcycle helmets. Int J crashworthiness. 6(1):53-64.

Hardy WN, Khalil TB and King AI (1994) Literature review of head injury biomechanics. International Journal of Impact Engineering 15(4): 561–568.

Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H., King, A. I., & Tashman, S. (2001). Investigation of head injury mechanisms using neutral density technology and high-speed biplanar X-ray. *Stapp carW.N. Hardy, et al. A study of the response of the human cadaver head to impact. Stapp Car Crash J, 51 (2007), p. 17 crash journal*, *45*, 337-368.

Hardy, W.N., Foster, C.D., Mason, M.J., Yang, K.H., & King, A. I. (2001). Investigation of head injury mechanisms using neutral density technology and high-speed biplanar x-ray. Staoo Car Crash Journal, 45, 337-368. Doi: 10.4271/2001-22-0016

Havlik, H. S. (2010). Equestrian sport-related injuries: a review of current literature. *Current sports medicine reports*, *9*(5), 299-302.

Havlik, Heather S. Equestrian Sport-Related Injuries, Current Sports Medicine Reports: September 2019 - Volume 9 - Issue 5 - p 299-302. doi: 10.1249/JSR.0b013e3181f32056

Ho, J., Zhou, Z., Li, X., & Kleiven, S. (2017). The peculiar properties of the falx and tentorium in brain injury biomechanics. *Journal of biomechanics*, *60*, 243-247.

Holbourn, A. H. S. (1943). Mechanics of head injury. *Lancet* 2, 438–441. doi:10.1016/S0140-6736(00)87453-X

Horgan, T. J., and M. D. Gilchrist. (2003). The creation of three-dimensional finite element models for simulating head impact biomechanics. Int. J. Crashworthiness 8:353–366.

Hoshizaki, T. B., Post, A., Kendall, M., Cournoyer, J., Rousseau, P., Gilchrist, M. D., Brien, S., Cusimano, M., and Marshall, S., 2017, "The Development of a Threshold Curve for the Understanding of Concussion in Sport," Trauma, 19(3), pp. 196–206.

Hoshizaki, T. B., Post, A., Oeur, R. A., & Brien, S. E. (2014). Current and future concepts in helmet and sports injury prevention. Neurosurgery, 75(suppl_4), S136-S148.

Hoshizaki TB, Brien SE. The science and design of head protection in sport. Neurosurgery. 2004;55(4):956-966; discussion 966-967.

Karton, C., & Hoshizaki, T. B. (2018). Concussive and subconcussive brain trauma: the complexity of impact biomechanics and injury risk in contact sport. *Handbook of clinical neurology*, *158*, 39-49.

King, A.I., Yang, K.H., Zhang, L., & Hardy, W. (2003). Is head injury caused by linear or angular acceleration? IROCDI Conference.

Kleiven, S. (2013) Why most traumatic brain injuries are not caused by linear acceleration but skull fractures are. Frontiers in Bioengineering and Biotechnology, 1, 15.

Kleiven, S. (2006) Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure, International Journal of Crashworthiness, 11:1, 65-79, DOI: 10.1533/ijcr.2005.0384.

Kuhl, H. N., Ritchie, D., Taveira-Dick, A. C., Hoefling, K. A., & Russo, S. A. (2014). Concussion history and knowledge base in competitive equestrian athletes. *Sports health*, *6*(2), 136-138.

Loyd, A. M. (2011). *Studies of the human head from neonate to adult: an inertial, geometrical and structural analysis with comparisons to the ATD head* (Doctoral dissertation, Duke University).

Loyd, A. M., Nightingale, R. W., Song, Y., Luck, J. F., Cutcliffe, H., & Myers, B. S. (2014). The response of the adult and ATD heads to impacts onto a rigid surface. *Accident Analysis & Prevention*, *72*, 219-229.

MacManus, D. B., Pierrat, B., Murphy, J. G., & Gilchrist, M. D. (2017). Region and species dependent mechanical properties of adolescent and young adult brain tissue. *Scientific reports*, *7*(1), 1-12.

Mattacola, C. G., Quintana, C., Crots, J., Tumlin, K. I., & Bonin, S. (2019). Repeated impacts diminish the impact performance of equestrian helmets. *Journal of sport rehabilitation*, *28*(4), 368-372.

McIntosh, A. S., Andersen, T. E., Bahr, R., Greenwald, R., Kleiven, S., Turner, M., ... & McCrory, P. (2011). Sports helmets now and in the future. British journal of sports medicine, 45(16), 1258-1265

McCrory P, Meeuwisse W, Johnston K, et al. Consensus statement on concussion in sport--the 3rd International Conference on concussion in sport, held in Zurich, November 2008. J Clin Neurosci Jun; 2009 16(6):755–763.

McCrory, P., & Turner, M. (2005). Equestrian injuries. *Epidemiology of Pediatric Sports Injuries*, *48*, 8-17.

Meaney, D. F., & Smith, D. H. (2011). Biomechanics of concussion. *Clinics in sports medicine*, *30*(1), 19–vii. https://doi.org/10.1016/j.csm.2010.08.009

MIPS® Safety System functionality. Mips. (2021, September 28). Retrieved October 4, 2021, from https://mipsprotection.com/science-technology/mips-safety-system-functionality.

Ommaya AK and Gennarelli TA (1974) Cerebral concussion and traumatic unconsciousness: correlation of experimental and clinical observations on blunt head injuries. Brain 97: 633–654.

Ommaya AK. Mechanical properties of tissues of the nervous system. J Biomech. 1968;1(2):127-138.

Ommaya AK, Hirsch AE. Tolerances for cerebral concussion from head impact and whiplash in primates. J Biomech. 1971;4(1):13-21.

Post, A., Blaine Hoshizaki, T., Gilchrist, M. D., and Cusimano, M. D., 2017. "Peak Linear and Rotational Acceleration Magnitude and Duration Effects on Maximum Principal Strain in the Corpus Callosum for Sport Impacts," J. Biomech., 61, pp. 183–192.

Post, A., Karton, C., Robidoux, M., Gilchrist, M. D., & Hoshizaki, T. B. (2019). An examination of the brain trauma in Novice and Midget ice hockey: Implications for helmet innovation. *CMBES Proceedings*, *42*.

Post, A., & Blaine Hoshizaki, T. (2015). Rotational acceleration, brain tissue strain, and the relationship to concussion. *Journal of biomechanical engineering*, *137*(3).

Post A, Hoshizaki TB. Mechanisms of brain impact injuries and their prediction: a review. Trauma. 2012;14(4):327-349.

Sae S. J211-1 instrumentation for impact test – part 1 – electronic instrumentation. SAE International; 2007. [AQ5].

Sengupta, P. (2013). The laboratory rat: relating its age with human's. *International journal of preventive medicine*, *4*(6), 624.

Stanfill, A. G., Wynja, K., Cao, X., Prescott, D., Shore, S., Baughman, B., ... & Tsao, J. W. (2020). Helmet use in equestrian athletes: opportunities for intervention. *Concussion*, (0), CNC85.

Test angles and impact locations. (2021, April 21). Mips. https://mipsprotection.com/science-technology/test-angles-and-impact-location/

Tiernan, S., & Byrne, G. (2019). The effect of impact location on brain strain. *Brain injury*, *33*(4), 427-434.

The test lab. (2021, April 26). Mips. https://mipsprotection.com/science-technology/test-lab/

Thomas, L., Roberts, V., Gurdjian, E. (1966). Experimental intracranial pressure gradients in the human skull. *Journal of Neurology, Neurosurgery and Psychiatry,* 29(5): 404-11

Thomas, L., Roberts, V., Gurdjian, E. (1967). Impact induced pressur4e gradients along three orthogonal axes in the human skull. *Journal of Neurosurgery,* 26(3): 316-21.

Trotta, A., Clark, J. M., McGoldrick, A., Gilchrist, M. D., & Annaidh, A. N. (2020). Biofidelic finite element modelling of brain trauma: Importance of the scalp in simulating head impact. *International Journal of Mechanical Sciences*, *173*, 105448.

Trotta, A., & Annaidh, A. N. (2019). Mechanical characterisation of human and porcine scalp tissue at dynamic strain rates. *Journal of the Mechanical Behavior of Biomedical Materials*, *100*, 103381.

Trotta, A., Zouzias, D., De Bruyne, G., & Ní Annaidh, A. (2018). The importance of the scalp in head impact kinematics. *Annals of biomedical engineering*, *46*(6), 831-840.

Turner, M., McCrory, P. and Halley, W. (2005), "Injuries in professional horse racing in Great Britain and the Republic of Ireland during 1992-2000", Br. J. Sports Med. Vol. 36, pp. 403–409.

Van Noort, R., Black, M. M., Martin, T. R. P., & Meanley, S. (1981). A study of the uniaxial mechanical properties of human dura mater preserved in glycerol. *Biomaterials*, *2*(1), 41-45.

Walker, A. E., Daniels, J. L., Weaver, N. L., and Robinson, P., 2000, "Jockey Injuries in the United States," J. Am. Med. Assoc., 283(10), pp. 1326–1328.

Wennberg, R., Tator, C. (2003). National Hockey League reported concussion. 1986-87 to 2001-02. *Canadian Journal of Neurological Science*, 30(3): 206-9.

Winkler, E. A., Yue, J. K., Burke, J. F., Chan, A. K., Dhall, S. S., Berger, M. S., ... & Tarapore, P. E. (2016). Adult sports-related traumatic brain injury in United States trauma centers. Neurosurgical focus, 40(4), E4.

Yoganandan N, Li J, Zhang J, et al. (2008) Influence of angular acceleration-deceleration pulse shapes

Zhang, L., Yang, K. H., and King, A. I., 2004. A Proposed Injury Threshold for Mild Traumatic Brain Injury. ASME J. Biomech. Eng., 126(2), pp. 226–236.

Appendix

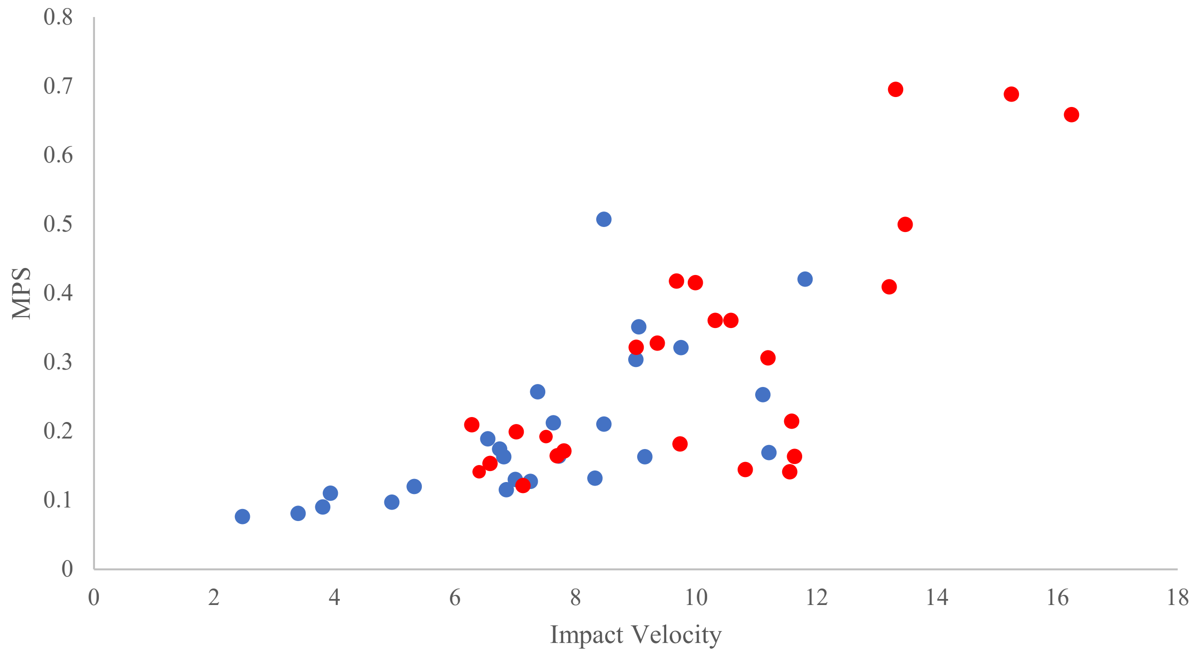

Figure 1. Mean reconstruction MPS result for each accident description as per Clark, Adanty, Post, Hoshizaki, Clissold et al, 2020.

Table 1

Average and ± standard deviation for low compliant impact results for the front, side, and rear boss impact locations.

Impact Location	Helmet Type	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/s^2)	Peak Rotational Velocity (rad/s)	Peak MPS
Front	No MIPS	103.77(1.89)	2256.1 (259.04)	8.13 (0.81)	0.404 (0.009)
	MIPS	106.6 (1.646)	2570.83 (124.05)	9.27 (0.91)	0.476 (0.004)

Side	No MIPS	136.4 (1.082)	3362.43 (348.82)	13.17 (4.77)	0.639 (0.074)
	MIPS	127.2 (8.107)	4120.23 (515.43)	14.77(2.06)	0.533 (0.039)
Rear Boss	No MIPS	111.2 (1.473)	5996.53 (832.53)	31.8 (1.31)	0.639 (0.074)
	MIPS	107.533 (2.715)	2920.27 (424.75)	17.83 (3.72)	0.342 (0.017)

Table 2

Average and ± standard deviation for high compliant impact results for the front, side, and rear boss impact locations.

Impact Location	Helmet Type	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/s^2)	Peak Rotational Velocity (rad/s)	Peak MPS
Front	No MIPS	54.467(1.387)	2819.53(922.602)	29.7(2.107)	0.264(0.0055)
	MIPS	47.433(1.415)	4004.2(101.328)	28.33(4.267)	0.25(0.008)
Side	No MIPS	61.65(2.568)	3070.73 (621.15)	22.85(6.807)	0.2715(0.0044)
	MIPS	61.033(0.902)	3021.633(197.539)	24.133(1.85)	0.3117 (0.008)
Rear Boss	No MIPS	51.967(0.723)	2992.167(421.892)	33.567(2.119)	0.231 (0.226)
	MIPS	49.267(0.321)	2727.933(194.12)	27.8(0.9)	0.226 (0.001)