THE INFLUENCE OF FULL FACIAL PROTECTION ON HEADFORM PEAK LINEAR ACCELERATION AT DIFFERENT HELMET IMPACT LOCATIONS

Leigh Jeffries[1], Carlos Zerpa[1], Eryk Przysucha[1], and Paolo Sanzo[1]

School of Kinesiology, Lakehead University, Thunder Bay, Canada[1]

The purpose of this study was to examine the influence of ice hockey helmet facial protection on measures of peak linear acceleration at different helmet locations during horizontal dynamic impacts. A repeated measures factorial ANOVA revealed a two-way interaction effect between type of facial shielding protection and helmet impact location on measures of peak linear acceleration, F (1.53, 26.05) = 18.60, $p < .001$, $\eta^2 = .52$. This finding indicates that the type of facial shielding protection can help mitigate linear impact accelerations at different helmet locations and consequently, minimize the risk of head injury. This outcome may have implications for athletes, coaches, and researchers to better understanding helmet performance in brain injury prevention for the sport of ice hockey.

KEY WORDS: ice hockey head injuries, helmet impact testing, facial shielding.

INTRODUCTION: Hockey has been identified as a sport with an elevated risk of concussions (Rowson, Rowson, & Duma, 2015). Concussion are a type of mild traumatic brain injury (mTBI) occurring from dynamic loading forces acting on a player's head or torso, which temporarily alter typical brain functions (Guskiewicz & Mihalik, 2006). Helmets are the primary form of head and brain protection used in ice hockey to minimize skull fractures with the hope of also minimizing concussions (Kis et al., 2013). The effectiveness of hockey helmets in preventing skull fractures and concussions was examined by measuring peak linear impact acceleration on different helmet locations during free falling testing protocols (Carlson, 2016; Walsh, Rousseau, & Hoshizaki, 2011). The outcome of these research studies revealed that hockey helmets responded differently to impacts at specifically defined helmet locations when reducing linear accelerations due to collisions. These differences in helmet properties at the different impact locations may affect helmet performance in protecting the head against concussions. Other researchers had suggested that the use of a hockey helmet with a facial shield protection made of metal or impact-resistant plastic might decrease linear impact accelerations of player's helmets; consequently, minimizing the risk of concussions (Graham, Rivara, Ford, & Spicer, 2014; Lemair & Pearsall, 2007). This approach seems controversial in the current literature. For example, Benson, Hamilton, Meeuwisse, McCrory, and Dvorak (2009) found no differences in concussion rates between players who wore facial protection to those who did not. Conversely, Lemair and Pearsall (2007), found that the use of facial shielding protection on a helmet reduced peak linear impact acceleration when the facial shield was impacted directly during a free falling collision test. Based on this gap in existing literature, the purpose of this study was to examine the influence of facial shielding protection on peak linear acceleration at different helmet impact locations during horizontal collisions. The dependent variable measured in this study was peak linear acceleration and the independent variables manipulated by the researchers were type of facial shielding protection and helmet impact location. It was hypothesized that the type of facial shielding protection would affect helmet impact location on measures of peak linear acceleration during horizontal collisions.

METHOD: A pneumatic horizontal impactor as depicted in Figure 1 was used to address the research hypothesis of this study. This horizontal impactor was designed and built at Lakehead University. Evidence of reliability and validity for the use of this impactor in helmet testing were provided in a research study conducted by Jeffries, Zerpa, Przysucha, Sanzo, and Carlson (2017). The horizontal impactor is composed of a main assembly, impactor arm, and linear bearing table. During data collection, the impactor arm was propelled into a helmeted medium sized NOCSAE headform at various velocities to simulate the mechanism of injury. The headform simulates the impact response of a 50th percentile adult human head. The headform contained an array of accelerometers to measure linear acceleration in the x, y, and z directions. The headform was mounted on a mechanical neckform designed to simulate the dynamic response of an adult human neck. The neckform was secured to a linear bearing table, which is adjusted to position the headform for impact testing at the front, front boss, side, rear boss, and rear locations (NOCSAE, 2016). For the current study, one medium sized hockey helmet with a dual density vinyl-nitrate liner was used to test three helmet/facial shielding conditions including helmet without facial shielding, helmet with a metal facial shield, and helmet with a polycarbonate facial shield.

Figure 1: Anterior and lateral views of the helmet/facial shield conditions fitted on a medium sized NOCSAE headform.

The helmet testing protocol for this study included 18 different impact velocities ranging from 2.01 to 5.13 m/s, which represented a range of head impacts speeds occurring in ice hockey. A calibrated compressed air tank propelled the impactor rod to strike the helmet at each impact locations for all 18 velocities. The sampling rate for data collection was 20 kHz for each accelerometer embedded in the headform. Each accelerometer analog signal was fed into a PowerLab analog to digital converter interfaced to a chart reader PowerLab software package. The software then computed the resultant acceleration by combining the acceleration in the x, y, and z directions into one vector. A low pass 1000 Hz filter was used to minimize noise due to any vibrations of the frame. The helmet was impacted for 15 combinations of impact location and facial shielding condition at 18 speeds, resulting in a total of 270 impacts. As degradation of the helmet was a concern, preliminary testing was conducted on one helmet to determine the number of impacts the helmet could withstand before increments of linear acceleration occur on the material properties of the helmet. This information was useful to help the researchers maintain the integrity of the data (Jeffries, 2017). Furthermore, the number of impacts administered in this study fall within the range of impacts that hockey players receive in a season (Brainard, et al., 2012). A 3 (facial shield conditions) x 5 (impact locations) two-way repeated-measures ANOVA was then conducted on measures of peak linear acceleration. One-way ANOVAs were used to help explain any interaction effect between facial shielding condition and helmet location. Bonferroni post hoc analyses were conducted for mean pair comparisons resulting from the one-way ANOVAs.

RESULTS: A significant two-way interaction effect between impact location and facial shielding condition was found on measures of peak linear acceleration, F (1.53, 26.05) = 18.60, $p < 0.001$, $\eta^2 = 0.52$. To help explain the interaction between these two independent variables as shown in Figure 2, one-way repeated measures ANOVAs were conducted and the results revealed significant differences across impact locations for no facial shielding, F (1.06, 17.98) = 31.83, $p < 0.001$, $\eta^2 = 65$; metal facial shielding, F (1.14, 19.37) = 31.06, $p < 0.001$, $\eta^2 = 0.65$; and polycarbonate facial shielding conditions, F (1.23, 20.84) = 20.90, $p < 0.001$, $\eta^2 = 0.52$. Bonferroni post hoc analysis revealed that the rear boss location received the highest and the rear location the lowest linear impact acceleration when compared to the other helmet impact locations for each facial shielding condition. The two-way interaction effect as shown in Figure 2 also revealed significant differences across facial shielding conditions for the front location F (1.07, 18.23) = 21.66, $p < 0.001$, $\eta^2 = 0.56$; front boss F (2, 34) = 29.60, $p < 0.001$, $\eta^2 = .64$; side location F (1.63, 27.75) = 4.01, $p = 0.037$, $\eta^2 = 0.19$; and rear boss location, F (1.29, 22.00) = 36.42, $p < 0.001$, $\eta^2 = 0.68$. Bonferroni post hoc analyses revealed that for the front boss and side locations with no facial shielding, the peak linear acceleration was higher than the metal and polycarbonate shielding conditions. For the rear boss location with metal shielding, however, the Bonferroni post hoc analysis revealed that the peak linear acceleration was higher than for the no facial shielding and polycarbonate conditions.

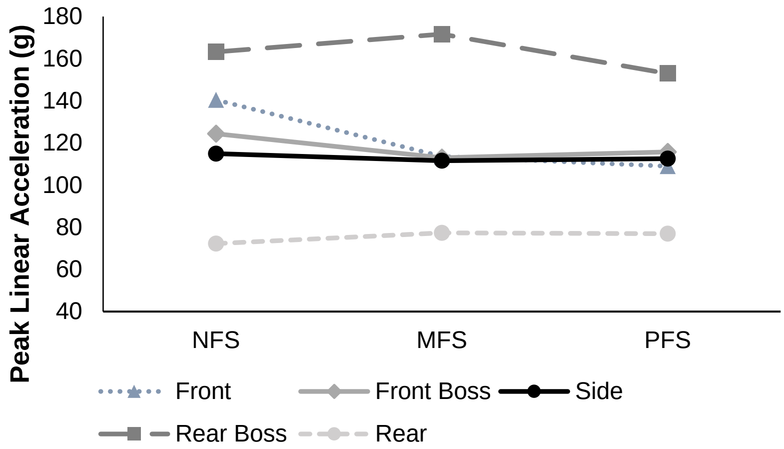

Figure 2: Impact locations and facial shielding conditions. Note that the facial shielding conditions on the horizontal axis are represented by grouping labels: no facial shielding (NFS), metal facial shielding (MFS), and polycarbonate facial shielding (PFS). Impact locations are represented by different lines.

DISCUSSION: The purpose of this study was to examine the influence of facial shielding on peak linear acceleration at different helmet impact locations during horizontal collisions. The results of this study revealed that helmets seem to perform differently in terms of reducing linear acceleration when impacted with and without facial shielding at different locations. The rear boss location, for example, produced the highest linear impact acceleration compared to the other locations. The rear location, on the contrary, produced the lowest linear impact accelerations. This outcome supports the research work of Carlson (2016) and Walsh, Rousseau, and Hoshizaki (2011), which stated that ice hockey helmets respond differently across impact locations in reducing linear accelerations due to collisions. Upon visual examination of the helmet, it was noted that the rear boss site had less attenuation lining than the other helmet locations, which may explain why this location had elevated acceleration values. The rear location, on the contrary, had the thickest lining. It was also possible that the geometry of the outer shell of the hockey helmet may have contributed to the differences observed between impact locations. Helmet shells tend to be geometrically "square" and have many external ridges; as a result, there are many flatter areas on the helmet which may affect its performance and ability to reduce peak linear accelerations (Halstead, Alexander, Cook, & Drew, 2000). In the current study, when a helmet had no facial protection, it was less efficient at reducing peak linear impact acceleration than when a helmet had metal facial shielding, and polycarbonate facial shielding. The increased ability of the helmet to reduce peak linear acceleration with facial shielding was observed at the front, front boss, and marginally at the side impact locations. This result supports the research work of Lemair and Pearsall (2007) who found substantial reductions in peak linear accelerations when directly impacting facial shields of helmeted headforms. Studies conducted in this area of research, however, seem to illustrate that certain areas of the helmet with or without facial protection are not as effective as others in reducing linear impact accelerations and, therefore, may be more associated with a higher risk of injury. When considering all impact sites of the helmet collectively, the outcome of this study indicates that the metal and polycarbonate facial shields were able to reduce the peak linear acceleration transferred to the headform from 123.04 g in the no facial shielding condition to 117.43 g and 113.42 g, respectively. The ability of the facial shield to reduce peak linear accelerations could be the difference between a hockey player sustaining or avoiding a concussion. This rationale can be supported by the work of Zhang, Yang, and King (2004) who determined that the probability of concussion for peak resultant linear accelerations at the centre of gravity of the head was estimated to be 66, 82, and 106 g for the 25th, 50th, and 80th percentiles indicating that as acceleration increases, there is a substantially increased risk of sustaining a concussion. The average accelerations in the current study were above these reported concussion probability values meaning that many of the impacts would have still resulted in concussions regardless of the facial protection condition. The small reductions in linear acceleration provided by wearing facial shielding, however, may help reduce the severity of the brain injury if a concussion was to occur.

CONCLUSION: This study found a two-way interaction effect between type of facial shielding protection and helmet impact location on measures of peak linear acceleration. This finding builds on the existing literature by adding information on the effect of facial shielding protection on peak linear acceleration across impact locations. The data analysis presented in the current study may provide an avenue for athletes and coaches to better understand hockey helmet performance in protecting the brain against concussions when wearing facial shielding protection. The outcome may also have implications for helmet manufacturers and researchers in the field of concussion.

REFERENCES:

Benson, B., Hamilton, G., Meeuwisse, W., McCrory, P., & Dvorak, J. (2009). Is protective equipment useful in preventing concussion? A systematic review of the literature. *British Journal of Sports Medicine*, *43*(1), i56-i67.

Brainard, L., Beckwith, J., Chu, J., Crisco, J., McAllister, T., Duhaime, A., ... & Greenwald, R. (2012). Gender differences in head impacts sustained by collegiate ice hockey players. *Medicine and science in sports and exercise*, *44*(2), 297-304.

Carlson, S. (2016). The influence of neck stiffness, impact location, and angle on peak linear acceleration, shear force, and energy loading measures of hockey helmet impacts. (Master’s thesis). Lakehead University, Thunder Bay, Canada.

Graham, R., Rivara, P., Ford, M., & Spicer, C., (2014). *Sports-Related Concussion in Youth.* Washington, DC: The National Academies Press.

Guskiewicz, K. M., & Mihalik, J. P. (2006). The biomechanics and pathomechanics of sports-related concussion. In S. M. Slobounov & W. J. Sebastianelli (Eds.), *Foundations of sports-related brain injuries* (65-80). New York, NY: Springer Science + Business Media, Inc.

Halstead, P. D., Alexander, C. F., Cook, E. M., & Drew, R. C. (2000). Hockey headgear and the adequacy of current designs and standards. In *Safety in Ice Hockey: Third Volume*. ASTM International.

Jeffries, L. (2017). The effect of facial protection, impact location, and neckform stiffness on peak linear acceleration, risk of injury, and energy loading measures of horizontal impacts on a hockey helmet impacts. (Master’s thesis). Lakehead University, Thunder Bay, Canada.

Jeffries, L., Zerpa, C., Przysucha, E., Sanzo, P., & Carlson, S. (2017). The use of a pneumatic horizontal impact system for helmet testing. *Journal of Safety in Engineering, 6*(1), 8-13.

Kis, M., Saunders, F. W., Kis, M., Irrcher, I., Tator, C. H., Bishop, P. J., & ten Hove, M. W. (2013). A method of evaluating helmet rotational acceleration protection using the Kingston Impact Simulator (KIS unit). *Clinical Journal of Sport Medicine*, *23*(6), 470-477.

Lemair, M., & Pearsall, D. J. (2007). Evaluation of impact attenuation of facial protectors in ice hockey helmets. *Sports Engineering*, *10*(2), 65-74.

NOCSAE. (2016). *Standard performance specification for newly manufactured ice hockey helmets.* (ND 030-11m16). Overland Park, USA: National Operating Committee on Standards for Athletic Equipment.

Rowson, B., Rowson, S., & Duma, S. M. (2015). Hockey STAR: A methodology for assessing the biomechanical performance of hockey helmets. *Annals of Biomedical Engineering, 43*(10), 2429-2433.

Walsh, E., Rousseau, P., & Hoshizaki, T. (2011). The influence of impact location and angle on the dynamic impact response of a Hybrid III headform. *Sports Engineering*, *13*(3), 135-143.

Zhang, L., Yang, K. H., & King, A. I. (2004). A proposed injury threshold for mild traumatic brain injury. *Journal of Biomechanical Engineering, 126*(2), 226-236.