International Journal of EXERCISE SCIENCE

*Original Research*

# Effect of Neck Strength on Simulated Head Impacts During Falls in Female Ice Hockey Players

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ABSTRACT

***International Journal of Exercise Science 14(1): 446-461, 2021.*** This study examined the effect of isometric cervical strength and impact location of the hockey helmet in mitigating the risk of concussions for two different mechanisms of injury from a fall during head impact simulation testing. Isometric cervical strength was measured on 25 female hockey players to compute and model neck strength on a mechanical neckform. A dual-rail vertical drop system with a helmet mounted on a surrogate headform simulated the mechanisms of injury causing concussions on female ice hockey players. Measures of peak linear acceleration and risk of injury due to a head collision (GSI) were used to assess the magnitude of the head impact due to a fall across three neck strength measures (weak, average, strong), three helmet locations (front, rear, side), and two mechanisms of injury (direct, whiplash+impact). A three-way ANOVA revealed a significant main effect for impact mechanism on the magnitude of peak linear acceleration and GSI, with the whiplash+impact mechanism generating significantly greater peak linear acceleration and GSI than the direct impact mechanism. A significant two-way interaction effect was found between impact location and mechanism of injury on peak linear acceleration measures, with the direct impact on the side location generating significantly greater peak linear acceleration than the frontal location. On the contrary, the whiplash+impact mechanism revealed that the frontal impact location produced significantly greater peak linear acceleration than the side location. This outcome suggests the geometry of the helmet material and the type of mechanism of injury both play a role in concussion risk.

KEY WORDS: Cervical, torque, injury mechanics, injury risk, whiplash, acceleration, concussion

## INTRODUCTION

Ice hockey is a fast-paced contact sport with inherent opportunity for injury due to high-speed shooting, low-friction ice surface, high acceleration-deceleration, and rapidly changing directions in skating (6). Ice hockey was once considered a male-dominated sport. Women's ice hockey participation rates, however, have changed that view as seen by the 900% increase in female participation over the past 15 years (15). The increase in women's participation has been paralleled with an increase in the calibre and competitiveness of female players. Consequently, despite rules prohibiting intentional body contact in women's hockey, injury rates of female hockey players have also increased (15). The most common injuries sustained by female hockey players are those involving the head and neck (16, 33), with concussions reported as the number one injury (1, 16). A concussion is a mild traumatic brain injury resulting from forces applied directly to the head or transmitted to the head from an indirect force (27). A concussion can cause acute and chronic physical, cognitive, and emotional symptoms that may negatively affect an individual's daily life (24).

Due to the growing concern surrounding traumatic brain injuries in the sport of ice hockey, the prevention of concussions has been identified as a research priority by the Centers for Disease Control and Prevention (13). The problem exists, however, in the limited research on concussions in female athletes. Evidence suggests that females may sustain concussions at a higher rate than males, experience more severe symptoms, and take longer to return to normal play (2, 4, 10). These concerning findings strengthen the need to improve the research involving female athletes. It remains unclear, however, which specific risk factors contribute to the heightened risk of concussions in females. Previous research examining concussions in women's hockey has primarily focused on the epidemiology of the injury (1, 20), while few studies have concentrated on the potential risk factors leading to concussions in a female-specific cohort (7, 38, 39). Of the research that has examined regarding the biomechanics of concussion risk in women's hockey, a few potential risk factors have been suggested such as anthropometric differences between males and females and biomechanical characteristics of head impacts in women's hockey (7, 38). Influential anthropometric differences may include measures of neck strength, neck circumference, and the ratio of neck circumference to head circumference since females are known to have weaker cervical muscle strength than their male counterparts (7, 22, 32). In addition, previous research identified female hockey players to experience significantly more impacts to the sides of the head than males, consequently increasing concussion risk (7, 37, 41).

Impact mechanisms causing concussions also seem to occur more frequently on female hockey players than males. Some researchers believe this outcome is because female hockey players experience less exposure to total ice time participation than males and consequently are less experienced in avoiding head collisions (7). Head impact mechanisms causing concussions in female hockey players, however, seem to happen more regularly from falls rather than direct collisions between female hockey players. The most common method of investigating biomechanics head collisions and falls in women's hockey has been using head impact telemetry (HIT) instrumented in helmets to measure the characteristics of the head impact during a real-time event. This method, however, is limited by the assumption that the helmet and skull move as a single rigid body (9). In addition, the HIT data needs to combine with synchronized video analysis to determine the nature of the head impact. As an alternative method, researchers use surrogate heads, mechanical neck-forms, and anvil impactors to investigate concussions and evaluate helmet performance via head impact simulations in a laboratory setting (5).

Peak linear acceleration as a measure of change in velocity over time provides an avenue to examine the magnitude of a head impact for different mechanisms of injury causing concussions in hockey players during falls to the ice or collisions between players (42). Zhang, Yang, and King (42) suggested that a peak resultant linear acceleration of 66 g, 82 g, and 106 g induced to the head during an impact corresponds to a 25%, 50%, and 80% probability of sustaining a concussion, respectively. These threshold levels resulted from typical head impact durations of 10 - 16 milliseconds. Peak linear acceleration in combination with the time duration of the impact also provides an avenue to assess the relative severity of a head injury resulting in concussions or skull fractures based on a set of tolerance criteria. The Gadd Severity Index (GSI), for example, offers a tolerance criterion in helmet testing, which is used by the National Operating Committee for the Standards on Athletic Equipment and also can be used in head impact testing examining concussions (29). Gadd (21) reported that an index of 1000 on the GSI represents the upper limit for severe brain injury. An index value generated at or near this level has a very highly probability of producing a concussion.

Research is being conducted to reconstruct injury mechanisms causing concussions via simulations, yet, needs to evolve to include specific anthropometrics or strength measures from specific target populations such as female ice hockey players. This approach will allow researchers to adjust the surrogate devices to achieve more realistic head impact biomechanics and, consequently, provide better information to develop methods or protective devices to mitigate the risk of concussions. Furthermore, simulation testing protocols may be limited by the inability of the technology to accurately reproduce real-life parameters.

Based on the limitations in existing research, this study aimed to:

a) Model isometric cervical strength data from a representative sample of female ice hockey players to adjust the stiffness of a mechanical neckform for dynamic testing of head impact biomechanics simulations.
b) Examine the effect of neck strength as a measure of torque on a mechanical neckform, head impact location, and mechanism of injury on measures of peak linear acceleration and injury severity index during simulated free-falling head impact testing while wearing a hockey helmet.

The research work was divided into three parts to address the purposes of the study: (I) human neck strength testing, (II) calibration of mechanical neckform, and (III) head impact simulation testing.

## METHODS

*Participants*

Part I: Human Neck Strength Testing: A total of 25 competitive female ice hockey players (age = 22.1 ± 2.6 years, playing experience = 15.8 ± 2.7 years, body mass = 71.7 ± 10.9 kg, height = 165.2 ± 5.2 cm) were recruited to participate in the isometric cervical muscle strength testing. A priori power analyses were conducted to estimate the participant sample-size based on pilot data related to measures of cervical strength using Equation 1 with a medium effect size of 0.5, a standard deviation of 17, and a power of rejection of 80% at $p < 0.05$ (12). The calculations indicated that a sample of at least 22 participants was required to achieve a power of rejection of 80% on measures of cervical strength.

$$N = 2^* S^{2*} (Z_\alpha + Z_B) / d^2 \quad (1)$$

Where:

N = sample size
S = the standard deviation obtained from previous pilot study
d = the accuracy of the estimate or how close to the true mean
$Z_\alpha$ = normal deviate for two- tailed alternative hypothesis at $p < 0.05$
$Z_B$ = power of rejection z score

The participants were members of the Lakehead University and Confederation College women's hockey teams and the Thunder Bay Women's Hockey Association (TBWHA) Senior House Division. Participants were included in the study if they had not been diagnosed with a concussion or other head/neck injury within the past six months to ensure that their ability to perform maximal isometric cervical contractions was not compromised. Any participant who sustained a head or neck injury within the past six months that prevented them from participating in their sport did not partake in the study unless they received medical clearance to return to play. Additionally, the participants had to be active players at the time of testing and had played at a caliber equivalent to or higher than their current caliber for the past three years to ensure that the sample was representative of the population (i.e. competitive female hockey players). Participants were considered healthy if they met the Physical Activity Readiness Questionnaire (Par-Q) form criteria and were free of any other musculoskeletal disorders that limited their ability to perform maximal isometric cervical contractions safely, as determined by a pre-screening questionnaire. Finally, to help reduce the risk of muscle strains during testing, participants were excluded if they had insufficient cervical range of motion based on the normal parameters identified by Swinkels and Swinkels-Meewisse (36). The study was approved by the institutional review board and all participants provided written consent prior to participation. This research was carried out fully in accordance with the ethical standards of the International Journal of Exercise Science (28).

*Protocol*

Part I: Neck Strength Testing: Maximal isometric muscle force production in cervical flexion, extension, and side flexion was measured during a single testing session lasting approximately 60 minutes. A Nautilus (Vancouver, WA, USA) four-way neck strength machine was chosen for the strength testing because it is a standard piece of equipment used for cervical muscle strengthening and has been used in previous research to examine maximal isometric cervical strength (8). The device was set up near a wall and bolted to the floor to ensure adequate stability. A strain gauge load cell, attached to the wall and connected perpendicularly to the moveable arm of the Nautilus machine via a lightweight chain, was used to measure the cervical muscle force of the participant produced during the maximal isometric strength tests. The lightweight chain caused the headrest to become static, providing a resistance to the participants' head when performing an isometric contraction.

A standardized dynamic warmup was performed before engaging in any testing to help minimize the risk of injury. This warmup included a five-minute stationary cycle at a moderate pace to increase core body temperature followed by dynamic neck stretches. A similar warmup protocol was included in previous research that required participants to perform maximal isometric cervical muscle strength testing (8). After the completion of the warm-up, neck range of motion was measured in flexion, extension, lateral flexion, and left and right rotation using a cervical range of motion (CROM) device (Performance Attainment Associates, Roseville, MN) to ensure that the participants were within the normal range for each movement direction (36). The CROM device has been used in previous clinical research to measure cervical range of motion (19, 30) and demonstrates high concurrent validity (3).

Total height, body mass, and the head and neck anthropometrics of the participants were also measured prior to testing. Neck length, neck circumference, and head circumference were measured with a measuring tape using the same anatomical landmarks as Seigler et al. (34). Neck length was measured from the occipital condyle to the C7-T1 point; neck circumference was measured around the laryngeal prominence; and head circumference was measured just above the level of the ears. These anatomical landmarks are comparable to those used in previous research (14). Head mass was estimated using the body segment parameter data from deLeva (17) and Zatsiorsky, Seluyanov, and Chugunova (40).

To measure isometric neck strength, the Nautilus neck strength machine was first adjusted to seat each participant for proper height so that the participant's forehead was positioned on the headrest with the horizontal metal bar at eye level. The participants were provided up to three practice trials at a submaximal effort to familiarize themselves with the testing protocol. For the test trials, they were instructed to steadily increase the amount of force applied over three seconds until they reached an isometric maximum. The maximum isometric force was held consistent for two seconds before relaxing (8). The participants were asked to choose their preferred side for the side flexion movements and the peak isometric force was applied at an angle of 10 degrees for each direction, to align with previous research (22). Verbal encouragement was provided for all trials and the peak force production, in Newtons, was measured by the load cell and captured by LabChart 7 software.

Participants performed three maximal isometric contractions each in flexion, extension, and side flexion (completed in that order) with a three-minute break between each trial to reduce the effects of muscle fatigue. The average maximum isometric force of the three trials was calculated for each movement direction. For the purposes of this study, the average overall cervical muscle strength was determined because the design of the mechanical neckform used during the head impact simulation testing only permits the modelling of overall cervical muscle strength. The researchers then computed the mean of the average maximal force in flexion, extension, and side-flexion. This method of developing an average overall cervical muscle strength was adopted from Collins et al. (14). Finally, based on the sample of data, the $10^{th}$, $50^{th}$, and $90^{th}$ percentile of average overall cervical muscle strength was calculated to represent a weak, average, and strong overall cervical muscle strength, respectively.

Part II: Calibration of Mechanical Neckform: A mechanical neckform constructed from neoprene rubber with steel discs to form the intervertebral discs of a human cervical spine was used to represent the 50th percentile of a human neck (35). The neckform contained a galvanized stainless-steel cable attached longitudinally through the center of the neck to adjust the stiffness of the rubber material. During the calibration process of the mechanical neckform, the cervical strength measures of force (N) from Part I were converted to measures of torque (Nm) to represent participants' neck strength levels. This process was accomplished by mounting a surrogate headform on the mechanical neckform and pulling on it using a single-axis strain gauge connected to a lightweight chain fixed to a steel frame. The surrogate head was instrumented with a CROM device to monitor a consistent angle of neck flexion of 10-degrees as participants during neck isometric testing flexed the neck approximately 10-degrees (Figure 1) (22). The stiffness of the neckform was adjusted to eleven different torques ranging from 0.84 Nm to 5.04 Nm in the flexed, extended, and laterally flexed directions. The mean of the three directions was then calculated to obtain an overall average force measure for each neckform torque to reflect the overall cervical muscle strength data gathered from the human participants. The measures of force and torque were then plotted and a linear interpolation equation was fitted to the data to estimate the torque required to adjust the stiffness of the mechanical neckform for a given force measure of cervical neck strength of a female hockey player participant. This process allowed the development of three neck torque values that corresponded to the 10th, 50th, and 90th percentile of overall neck strength for the sample of competitive female hockey players, which ultimately were used to represent a weak (1.36 Nm), average (2.94 Nm), and strong (4.62 Nm) neck stiffness during the head impact simulation testing of Part III of this study.

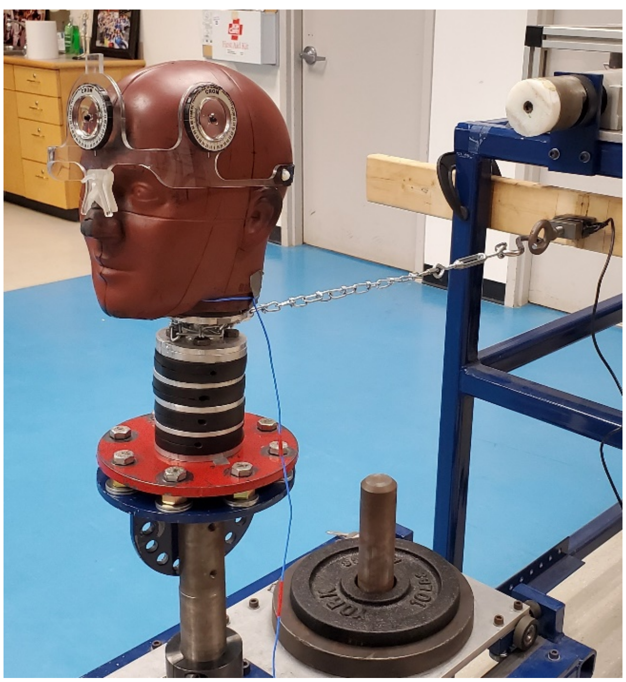

**Figure 1.** Setup of the mechanical neckform calibration.

**Figure 2**. Setup of the head impact simulation testing.

Part III - Head Impact Simulation Testing: A medium-sized National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform connected to the mechanical neckform simulated the free-falling head impact testing. The NOCSAE surrogate headform accurately features the anatomical bone structure and facial characteristics of a human head (26). The NOCSAE headform used in this study represented the dimensions of a 50th percentile adult head and it was instrumented with an array of triaxial accelerometers to measure the linear acceleration of the head in the anterior-posterior, superior-inferior, and the left-right directions (26). A medium-size hockey helmet was mounted on the headform to conduct the head surrogate impact testing. The helmet also featured a full cage facial shield due to the equipment regulations of women's hockey.

A dual-rail free-falling impact system was used to simulate the head impact mechanisms of injury (Figure 2). The system has strong evidence of reliability (ICC = 0.922, $p < 0.005$) and strong evidence of concurrent validity on measures of peak linear acceleration (ICC = 0.844-0.952, $p < 0.005$) when compared to a standardized NOCSAE drop system from the University of Ottawa Neurotrauma Science research lab, instrumented with a hybrid III neckform and NOCSAE headform (11). The free-falling impact system featured a drop carriage with the neckform and headform mounted on it. The drop carriage attached to two vertical rails of the system and moved along the rails with little friction, allowing a free-fall drop to occur onto a steel plate anvil. The position of the headform in the carriage was adjusted to control which location of the head sustained the impact when the drop carriage was released. The combined weight of the headform, neckform, hockey helmet, and drop carriage was 30.6 kg and this weight remained consistent throughout the testing.

The drop simulations were performed in accordance with the NOCSAE drop test standards protocol. According to these standards, the headform, equipped with properly fitted headgear, is to be positioned in the drop carriage and dropped from a desired height to reach desired freefall velocity (29). The NOCSAE protocol was chosen for the current study because this protocol is designed to provide reliable and repeatable measurements of linear acceleration experienced by a surrogate headform when testing hockey helmets (29). A hockey helmet was mounted on the headform for all simulated head impacts and the helmet was replaced with an identical new helmet after every 90 impacts. At impact, the instantaneous linear acceleration of the head was measured via triaxial accelerometers instrumented in the headform sampling at a frequency of 20 KHz. A low pass SAE j211 filter with a cut-off frequency of 1000 Hz eliminated the noise generated due to vibrations induced to the headform during free falling. The resultant acceleration (Equation 2) was used to combine the acceleration measures in the three directions and calculate GSI (Equation 3), aligning with NOCSAE standards (29):

$$\text{Resultant Acceleration} = \sqrt{x^2 + y^2 + z^2} \tag{2}$$

Where:

x = linear acceleration in the x-direction
y = linear acceleration in the y-direction
z = acceleration in the z-direction

$$\mathrm{GSI} = \int_{t0}^{t1} A^{2.5} dt \quad (3)$$

Where:

A = head acceleration impulse function
$t_1$ = impulse duration
dt = sampling time

The headform was dropped from 16 different heights resulting in 16 different impact speeds ranging from 2.62 to 4.64 m/s, similar to a head drop protocol used in previous research to simulate players' impacts at different freefall velocities (12). One trial was performed for every combination of the three neck stiffnesses (weak, average, strong), three impact locations (front, rear, and side), and two impact mechanisms (direct and whiplash+impact) across 16 speeds, resulted in a total of 288 impacts.

*Statistical Analysis*
Descriptive statistics were used to calculate the $10^{th}$, $50^{th}$, and $90^{th}$ percentiles of the cervical muscle strength in flexion, extension, and side flexion, as well as overall cervical muscle strength. Descriptive statistics were also used to calculate the $10^{th}$, $50^{th}$, and $90^{th}$ percentiles of head mass, head circumference, neck circumference, and neck length.

To examine the effect of neckform torque, head impact location, and impact mechanism on peak linear acceleration and GSI, inferential statistics were used to analyze each dependent variable separately. A $3_{(\text{neckform torque})} \times 3_{(\text{head impact location})} \times 2_{(\text{impact mechanism})}$, completely randomized, factorial ANOVA was conducted to examine the interaction effect of these three factors on measures of peak linear acceleration and GSI at an alpha level of $p \leq 05$. If there were no significant three-way interaction effect observed, the significant main effects of the independent factors were analyzed for each dependent variable separately. Next, two-way ANOVAs were conducted to further analyze any significant two-way interaction effects among independent factors on each dependent variable separately. Finally, a series of one-way ANOVAs were conducted to explain the simple main effects observed within the levels of each independent factor. Since neckform torque and impact location were defined by three levels, Bonferroni post-hoc analyses pair comparisons were used for each of the dependent variables.

**RESULTS**

Part I: Neck Strength Testing: Since the overall cervical muscle strength data exhibited a normal distribution pattern and participants' data fell within the $10^{th}$, $50^{th}$, and $90^{th}$ percentiles, the researchers calculated the head and neck anthropometric measures within these percentiles to provide a set of normative data characterizing the cervical muscle strength of female hockey players who participated in the current study.

Although Part III of the study only utilized the $10^{th}$, $50^{th}$, and $90^{th}$ percentiles of overall cervical muscle strength, the researchers also calculated the $5^{th}$ and $95^{th}$ percentiles to provide a more robust set of data of head and neck anthropometric measures for future research work. The results of the cervical muscle strength testing are summarized in Table 1 along with head and neck anthropometric measures.

**Table 1.** Cervical muscle strength and head and neck anthropometric measures (percentile rank and SD) for a sample of female ice hockey players ($n$ = 25).

| Measures | 5th | 10th | 50th | 90th | 95th | SD |
|---|---|---|---|---|---|---|
| Flexion (N) | 43.67 | 54.27 | 72.62 | 91.82 | 101.20 | 14.94 |
| Extension (N) | 53.56 | 59.67 | 95.00 | 136.73 | 153.49 | 25.12 |
| Side Flexion (N) | 53.74 | 56.17 | 76.47 | 98.66 | 126.66 | 19.24 |
| Overall (N) | 51.37 | 58.64 | 76.01 | 108.27 | 121.16 | 17.52 |
| Neck Length (cm) | 6.6 | 7.0 | 8.3 | 9.2 | 9.8 | 0.7 |
| Neck Circumference (cm) | 30.7 | 31.4 | 34.0 | 36.1 | 36.6 | 1.7 |
| Head Circumference (cm) | 52.5 | 53.5 | 56.1 | 58.8 | 59.7 | 2.0 |
| Head Mass (kg) | 3.6 | 3.9 | 4.8 | 5.8 | 6.5 | 0.7 |

Part II: Calibration of Mechanical Neckform: The neckform calibration procedure generated a linear relationship between neckform torque and overall force, which accounts for 96.88% of predicted variance leaving only 3.12% of variance not accounted for by this model. The linear relationship between neck torque and force can be defined by Equation 4 and observed in Figure 3 to compute the amount of neck torque required to adjust the stiffness of the mechanical neckform for a given force measure of participant neck strength.

$$y = 0.0213x - 4.2881 \tag{4}$$

Where:

x = force (N) and y = torque (Nm)

Part III: Head Impact Simulation Testing: The results revealed no significant three-way interaction effect among neckform torque, head impact location, and impact mechanism on measures of peak linear acceleration, $F(4, 270) = 0.50$, $p > 0.05$. There was, however, a statistically significant main effect of impact mechanism on peak linear acceleration with a small effect size, ($F(1, 270) = 55.60$, $p < 0.05$, $\eta^2 = 0.17$). This revealed that the whiplash+impact mechanism (M = 143.86 g, SD = 29.42 g) generated significantly greater peak linear acceleration than direct impacts (M = 115.98 g, SD = 35.38 g).

Results also revealed a significant two-way interaction effect between impact location and impact mechanism with a small effect size, $F(2, 270) = 10.40$, $p < 0.05$, $\eta^2 = 0.07$ (Figure 4). Upon further investigation of the two-way interaction effect between impact location and impact mechanism, significant differences in peak linear acceleration among impact locations were found for the direct impact mechanism, $F(2, 141) = 5.90$, $p < 0.05$, $\eta^2 = 0.08$. A Bonferroni post hoc comparison revealed that side impacts (M = 126.13 g, SD = 34.12 g) resulted in significantly higher levels of peak linear acceleration than frontal impacts (M = 102.74 g, SD = 39.24 g) during direct head impacts, $p < 0.05$.

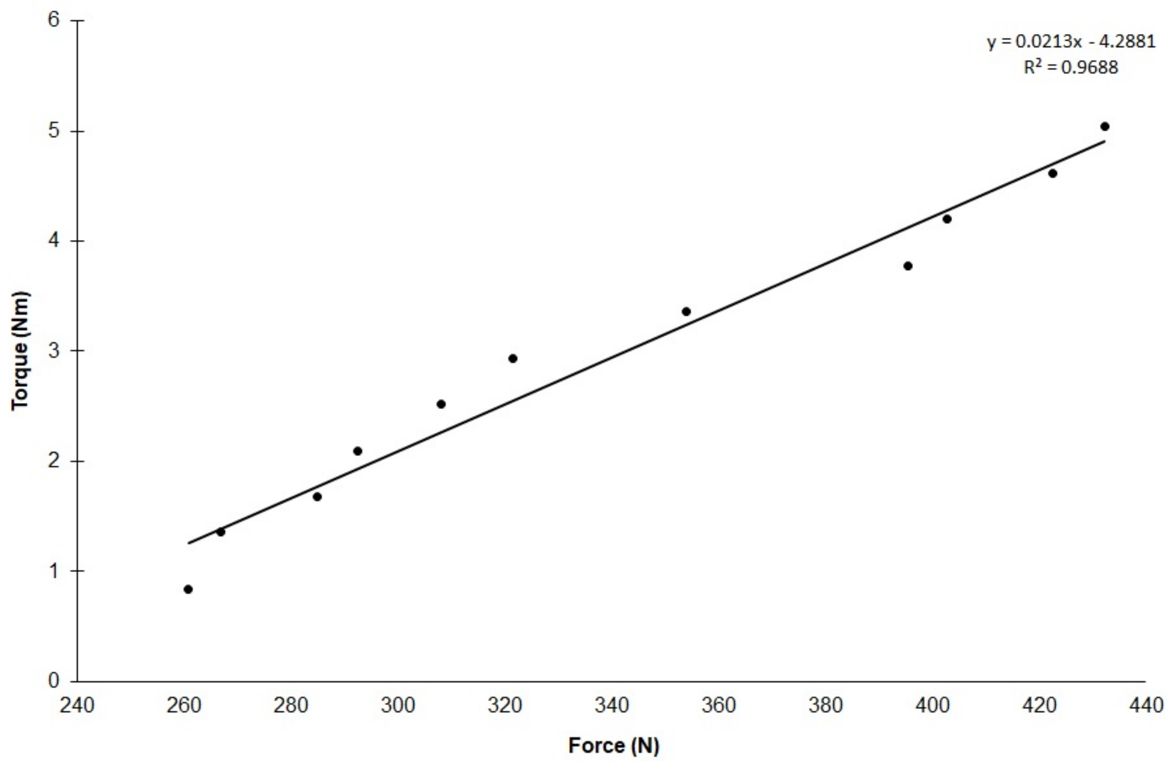


**Figure 3**. The relationship between torque and force for the mechanical neckform.

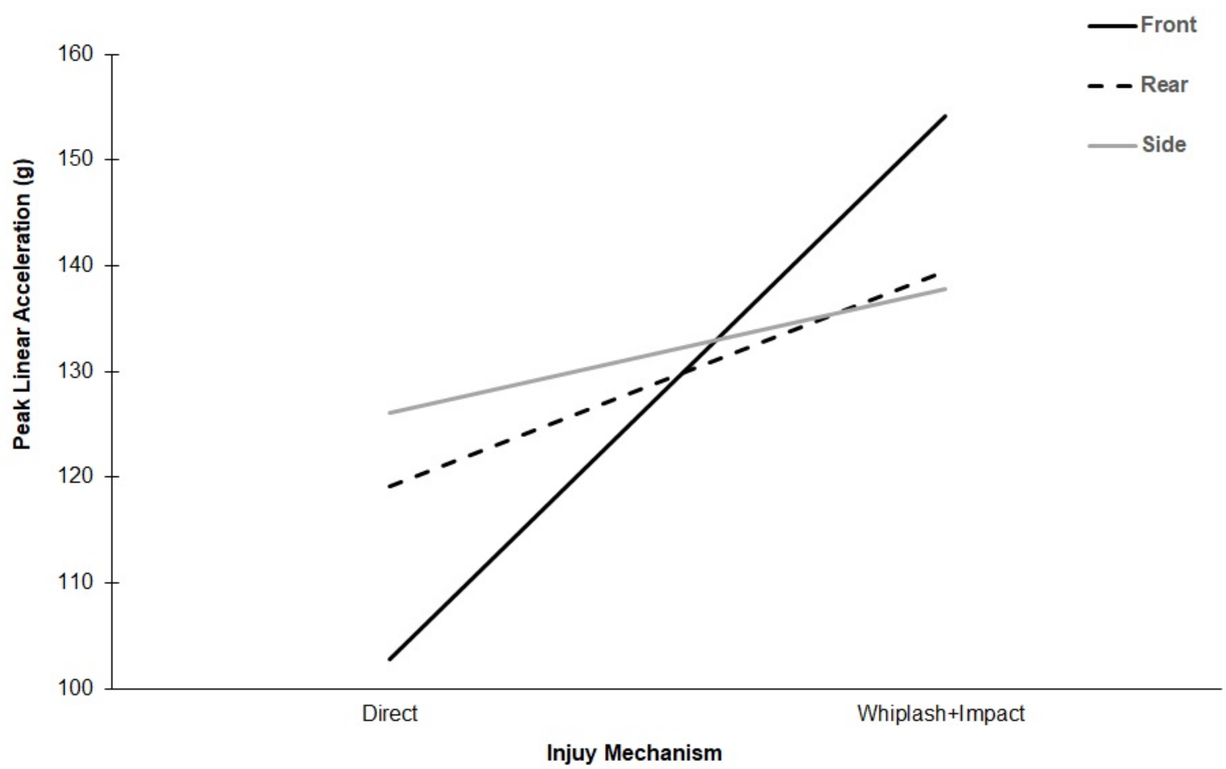


**Figure 4**. Two-way interaction of impact mechanism and impact location on peak linear acceleration. Linear acceleration was greater during the whiplash+impact mechanism for all impact locations.

A significant simple main effect for peak linear acceleration among impact locations for the whiplash+impact mechanism was also observed, ($F(2, 141) = 4.73$, $p < 0.05$, $\eta^2 = 0.06$). A Bonferroni post hoc analysis revealed significantly higher peak linear acceleration during frontal impacts (M = 154.21 g, SD = 4.85 g) compared to rear (M = 139.50 g, SD = 35.79 g) and side (M = 137.86 g, SD = 23.85 g) impacts for the whiplash+impact mechanism.

Finally, the results revealed no significant three-way interaction effect among neckform torque, impact location, and impact mechanism on measures of GSI, ($F(4, 270) = 0.35$, $p > 0.05$). Results did, however, demonstrate a significant main effect of impact mechanism on GSI, ($F(1, 270) = 68.18$, $p < 0.05$, $\eta^2 = 0.20$). The direct head impacts produced significantly lower mean GSI (M = 405.96 g, SD = 217.76 g) than the whiplash+impact mechanism (M = 550.44 g, SD = 328.50 g).

## DISCUSSION

Concussions in women's hockey are a significant concern, yet the underlying risk factors remain unclear. This research aimed to combine human data with simulation-based head impact testing to explore risk factors related to concussions in a female-specific population. This research was divided into three main parts: the neck strength testing and anthropometric measurements; the neckform calibration and the head impact simulation testing.

One of the goals for the neck strength testing was to develop a set of normative data describing the cervical muscle strength and anthropometric measures of a sample of competitive female hockey players to establish a more robust dataset upon which further research may be positioned. It was also necessary to confirm the ecological validity of the head impact testing to the target population. Although, epidemiological studies have found that females have lower cervical muscle strength than males, this outcome appears to be consistent when comparing the isometric cervical strength of the current sample of female hockey players to a sample of male hockey players who were tested using similar equipment and protocols (8, 22, 33). These findings have significant implications because cervical muscle strength is a factor that many researchers suggest is linked to concussions in athletes, including female hockey players (7, 14, 18). Stronger cervical musculature may have the ability to mitigate external forces applied to the head, thereby reducing the risk of sustaining a concussion from an impact (18). Unfortunately, there is limited cervical muscle strength data in current research specific to female hockey players and therefore, it becomes important to develop normative data documenting the cervical muscle strength of female hockey players to use in future research.

In addition to cervical muscle strength, previous research has identified the potential role of head and neck anthropometry in concussions (14). It has been suggested that athletes with a smaller neck circumference to head circumference ratio, smaller neck circumference, and a greater head mass may be more susceptible to concussions due to the decreased ability to mitigate biomechanical forces applied to the head during an impact (14). When comparing the female data of the current study to that of Broennle et al. (8) who measured head and neck anthropometrics of male hockey players, females appeared to have a smaller mean neck circumference (M = 34.0 cm, SD = 1.7 cm) and neck length (M = 8.3 cm, SD = 0.9 cm) than males (neck circumference: M = 39.0 cm, SD =1.6 cm; neck length: M = 11.9 cm, SD = 1.4 cm), raising the concern regarding the potential role of head and neck circumference in the heightened concussion risk for female athletes. Therefore, since existing anthropometric data specific to female hockey players is scarce and the current research did not examine differences in anthropometric measures between male and female hockey players, the data collected in the current study can also be used in future research exploring the relationship between anthropometric measures and concussions in the sport of ice hockey for males and females.

Concussion research in athletes, however, typically assumes one of two forms: on-field assessment using real-time data or simulated reconstruction of head impacts. To our knowledge, this is the first study to model human strength measure on a mechanical neckform during head impact simulation testing for female ice hockey players. By using human data to adjust the stiffness of the mechanical neckform, the results of the study, specifically pertaining to the influence of neckform torque, can be more confidently generalized back to female ice hockey players. Moreover, since this was the first study to apply human strength and anthropometrics to surrogate devices for a population of female ice hockey players, the outcome of this research builds on the implementation of human data to allow for the application of concussion simulation-based research.

The main outcomes related to the head impact simulation testing revealed that neckform torque did not have a significant effect on head impact biomechanics, although impact mechanism and head impact location did have a significant effect on outcome measures. Specifically, the whiplash+impact mechanism produced significantly greater peak linear acceleration than direct impacts, suggesting that perhaps other, more common impact mechanisms found in women's hockey, including a combined whiplash+impact mechanism, are increasing the players' risk for sustaining a concussion. More attention should be paid to impact mechanisms that result in a combined inertial loading and direct impact loading experienced by the head in an effort to help characterize the risk of these mechanisms in real life. Furthermore, side head impacts appeared to generate significantly greater amounts of peak linear acceleration than frontal impacts. These results are comparable to previous simulation-based concussion research (37), as well as initial animal studies (23). That is, side impacts have been found to result in greater levels of acceleration than other impact locations and may also experience more intercranial tissue damage, potentially leading to greater concussion risk (41). Furthermore, full-facial cages have the potential to alter the amount of force applied to the head during impact due to the altered geometry of the helmet. Lemair and Pearsall (25) revealed that full cages significantly reduce the peak linear acceleration experienced by the head during a direct impact. This outcome is likely attributed to the ability of the cage to distribute some of the forces radially, away from the head's center of mass. The chin support in full facial shields is also speculated to reduce forces experienced by the head (25). The full cage used in the current study, therefore, likely played an important role in reducing the peak linear acceleration in the frontal impacts.

Finally, the whiplash+impact mechanism resulted in significantly greater measures of severity index than the direct impact mechanism, which was to be expected due to the significantly higher peak linear accelerations observed during the whiplash+impact mechanism. Moreover, the impact duration was also greater during the whiplash+impact mechanism as compared to the direct impacts. Since calculations of GSI are based on both peak linear acceleration and impact duration, it was logical to observe significantly greater GSI measures during the whiplash+impact mechanism. This finding only reinforces the need to explore impact mechanisms other than direct head impacts, due to the potential injury risk that they pose to athletes.

This study was limited by the inability to measure angular acceleration of the head due to the configuration of accelerometers within the headform. Due to the strong link between angular acceleration and concussion, using a headform with the ability to measure angular acceleration would further strengthen the data gathered in future simulation testing (31). From the practical perspective, the results of this study indicate that impact mechanisms other than direct head impacts have the potential to cause concussions, which emphasizes the need for future research to investigate other injury mechanisms. Additionally, impact location appeared to be influential in injury risk. Coaching staff and players should be aware of the potential risk associated with impacts to sides of the head, as well as the potential protective factor that full facial shields, specifically cages, have in the mitigation of concussive forces for female ice hockey players. Finally, neckform torque did not appear to influence concussion risk, although players should not rule out the possible benefit of having strong neck muscles to help oppose external forces during impacts.

Future research should continue to incorporate human data in simulation-based concussion research to improve the mechanical neck designs and gain a better understanding of the behaviour of the head and neck during impacts as it applies to specific target populations. This information will have implications for researchers, coaches, and athletes since there is empirical evidence to suggest the potential link between cervical muscle strength and concussions (14, 18).

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