be found in Figs. 4 and 5. Significant main effects were found for all comparisons ($p<0.05$). The youth and adult collisions had significantly lower magnitudes of response in comparison to the falls. In comparison between the adult and youth responses, no significance was found between rotational acceleration and MPS for the fall impacts between these two groups ($p>0.05$). In addition, the rotational accelerations were not found to be significantly different between the adult and youth collision impacts ($p>0.05$). All other comparisons between the adult and youth responses were significantly different ($p<0.05$).

TABLE III
COMPARISON OF RESPONSES BETWEEN THE ADULT AND YOUTH FALL IMPACTS

		Peak resultant acceleration		
	Velocity (m/s)	Linear (g)	Rotational (rad/s^2)	MPS
Youth	3.8	83.3 (8.0)	6198.7 (2665.4)	0.435 (0.12)
Adult	4.47	131.4 (18.2)	5868.6 (2114.3)	0.404 (0.13)

TABLE IV
COMPARISON OF REPSONSES BETWEEN THE ADULT AND YOUTH COLLISION IMPACTS

		Peak resultant acceleration		
	Velocity (m/s)	Linear (g)	Rotational (rad/s^2)	MPS
Youth	3.0	14.7 (0.7)	1269.4 (257.1)	0.143 (0.02)
Adult	6.5	32.5 (5.3)	2661.4 (653.5)	0.256 (0.04)

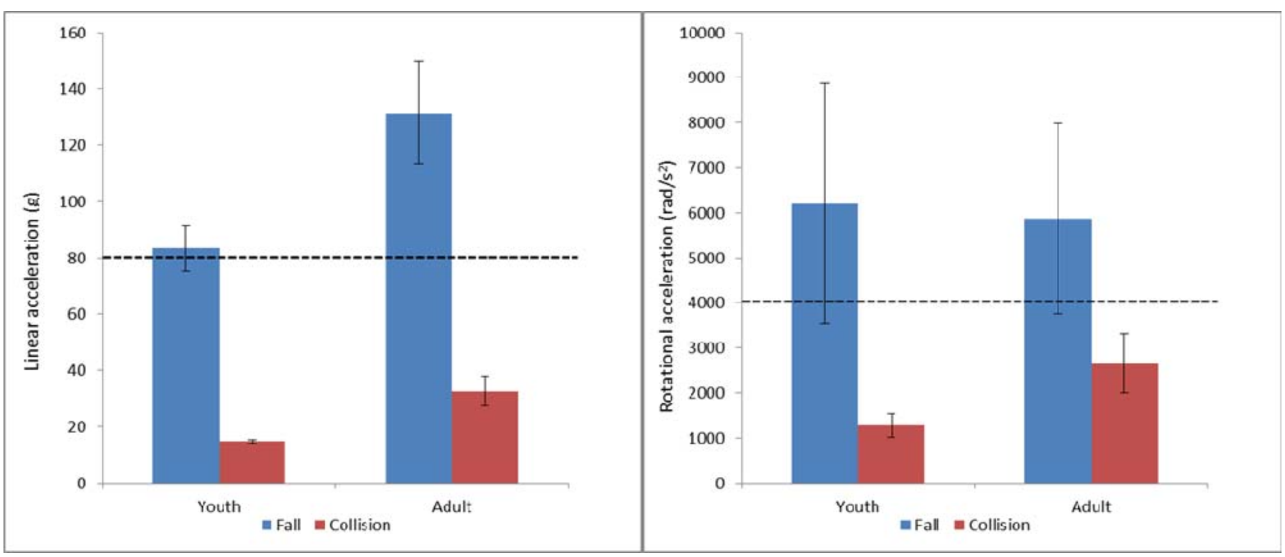

Fig 4. Linear (left) and rotational (right) acceleration comparisons for the fall and collision impacts. Black dotted line denotes magnitudes above which brain injury has been reported to occur in the literature for adults [11-12][43-46]

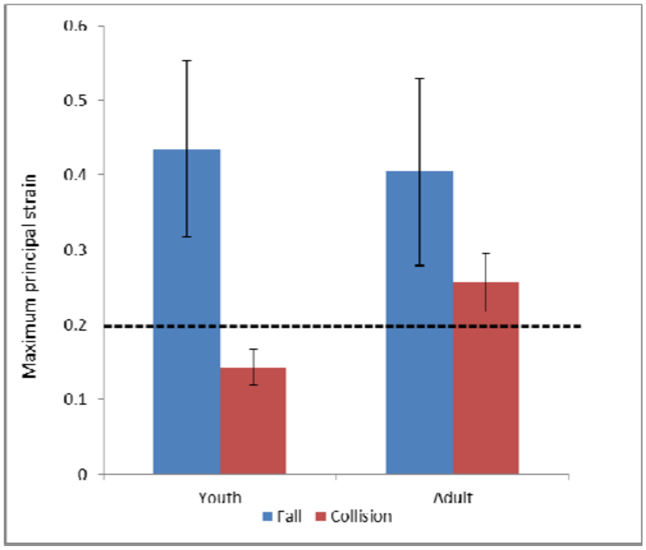

Fig 5. Maximum principal strain comparisons for the fall and collision impacts. Black dotted line denotes magnitudes above which brain injury has been reported to occur in the literature for adults [11-13][47].

IV. Discussion

Significant differences were found between the adult and youth for the helmeted headform impact conditions. When examining just the adult impact condition, there was a risk of brain injury reflected by the linear and rotational acceleration and MPS magnitudes for the falling impacts, whereas risk was identified through only MPS for the collision impacts (Figs 2 and 3). This may be a reflection of the longer duration of the linear and rotational accelerations of the collision events, which were typically around 20-25 ms. These long duration acceleration pulses result in strains in the brain tissues at lower magnitudes of peak acceleration. This phenomena has been described previously by [48] through impacts using primate subjects, demonstrating that concussion often occurred through longer duration acceleration pulses. Research conducted by [49] and [50] have also identified this relationship, as well as [51] using finite element and computational methods. The youth impacts demonstrated that for the parameters that describe the typical falling and collision events that the risk of brain injury would be primarily from falls to the ice. While the collision events were long duration accelerations to the brain (20 ms), the low magnitudes (14.7 g and 1.3 krad/s^2) were not sufficient to create a damaging strain environment (0.14 MPS) within the model. This may be a reflection of the lower velocities and masses that are involved in the collision events for youth ice hockey players [14]. It is likely that the conditions of the impact event do not create enough of a transfer of energy to the helmet and head to create a high risk of brain injury. This result coincides with research involving concussion in youth ice hockey that identified falls to the ice as being the primary cause of brain injury [2]. As this is a frequent event for youth playing ice hockey, future helmet designs should reflect fall to ice impact events under youth impact conditions to aid in the improvement of helmet designs. The improvement in impact absorbing capacity could be gained from adjusting the density of the impact absorbing liner to perform within the youth ranges of energy of impact. Making adjustments to the shell may also be beneficial; however improvements of the liner would likely be the easiest route with the largest effect. In addition the current ice hockey helmet standard would need modification to account for the energies that the youth players are encountering when a brain injury is likely to occur. For adults, ice hockey helmet development and standards may consider focusing on collision events in addition to falls, as collisions have been reported as the event most likely to result in concussion for that age group [6]. This improved protection for adults may reside in a stiffer liner for falls, and possible a secondary rotational acceleration damping technology to help mitigate injury from shoulder contacts.

This research should be interpreted with certain limitations. The tolerance literatures used as points of comparison in this research are from human adult reconstructions. It is possible that the tolerances to brain injury for youth may differ somewhat from the adult magnitudes. To date there is no human youth concussion threshold data available with which to compare this data to, which is why adult data was used to help put the results in context. The headforms are commonly used for impact research, and while they have been described to produce results in the range of cadaveric testing may not provide completely biofidelic responses [52]. The impact parameters were determined through measures from literature sources and are representative of the impacts for the target age groups but do not represent actual brain injury reconstruction. The results are specific to those assumptions and representations from the literature of impact mass, velocity, compliance, and location. In addition, the UCDBTM and scaled version are both dependent on the material properties and conditions that define the interactions of the represented tissues and all results from these models should be considered in light of those limitations.

V. CONCLUSIONS

This research identifies adult ice hockey players are likely at risk of concussion from both collisions and falls, whereas youth are at risk primarily from falls. These results suggest that to improve protection from brain injuries for youth that the helmet liner density should be adjusted to maximize the energy absorbing capacity of the helmet for falls. For adults, it is likely that improvements of helmet design needs to focus on both falls and collision conditions for improved protection for concussion. New helmet designs may need to focus on dual energy absorbing systems, one for linear falls, and the other for rotational impacts that are common to shoulder to head contacts. In addition, to improve protection new brain injury tolerances specific to youth must be researched to determine appropriate thresholds to target to optimize helmet designs and standard development.

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