Human factors and traumatic injury considerations associated with small changes in combat helmet mass

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Abstract. Reducing the mass of combat helmets will benefit the user's comfort and mobility and decrease the risk of musculoskeletal injury. Technological advancements have resulted in small incremental reductions in helmet mass. However, there is limited data quantifying the potential user benefits associated with these small mass reductions. This research considered both human factors and traumatic injury potential associated with 200 g changes in helmet mass. The human factors assessment comprised three helmets of different masses (1.0, 1.2 and 1.4 kg) worn by 20 Army qualified riflemen and applied a within-subject design. A carbon 'bump' helmet was modified to allow mass to be discretely added to the crown of the helmet while participants remained blinded to the mass condition. After completing two hours of representative military tasks, a questionnaire concerning the human factors effects of helmet mass was completed. The risk of neck injury during frontal vehicle impacts was determined via validated male and female MADYMO models for five different helmet masses ranging from 0.6 to 1.4 kg in 200 g intervals and for a no helmet condition. The helmet moment of inertia was scaled appropriately with the mass and the crash pulse consistent across all simulations. The feedback from participants suggested a 200 g helmet mass reduction is likely to a have a small positive impact on human factors. Similarly, the neck injury risk during frontal impacts is only slightly reduced with a 200 g change in mass. This reduction in injury risk was equivalent to a reduction in impact speed of between 1.3 and 5.2 km/h for males and 2.7 and 5.3 km/h for females for driving speeds 20 to 80 km/h. Although a 200 g reduction in helmet mass was found to induce only small effects, any reduction is head supported mass is seen as beneficial.

1. INTRODUCTION

1.1 Human factors considerations

The addition of head supported mass (HSM) causes the centre of mass (COM) of the head/HSM system to move away from the head's natural COM as well as increasing the head/HSM system mass moment of inertia (MMOI) and the resultant torque on the spine when moving the head. Ivancevic and Beagley provided three basic ergonomic recommendations for choice of HSM [1]. These were to choose an HSM with the smallest mass, with the mass most aligned to the head's natural COM, and with the mass closest to the head in terms of smallest diameter to limit MMOI [1].

The increased muscle activity required to support HSM as well as to maintain the head in an upright position to counter the effects of head-helmet system COM changes induced by the HSM can cause muscle fatigue [2-4]. Such muscle fatigue can cause pain and soreness which can cause distraction and impact operational performance [5]. The strain and increase in fatigue associated with increased HSM has been found to cause reduced efficiency, lower mental concentration and to increase the potential to make mistakes [6], as well as a decrease in the speed and number of head scanning movements [7]. Increased MMOI causes similar issues relating to neck muscle fatigue and the indirect effects these can have on operational performance. Physical discomfort has been found to be affected by HSM with a linear relationship between reports of discomfort and an increase in HSM [7, 8]. Discomfort can cause headaches and distraction which can subsequently affect operational performance [6, 9, 10]. MMOI arguably has the most direct effect on operational performance. Increased MMOI decreases head angular accelerations (for starting and stopping motion) and can therefore cause delayed head movements which can result in deficiencies in performance of tasks including those requiring efficient tracking and sighting of a moving target [11].

The mass of a helmet system has been proposed as a factor that contributes to overall user acceptance [12]. User acceptance is a multifaceted attribute and failure to achieve user acceptance can result in misuse and disuse of an item of equipment [6]. One of the key issues is that there is very limited data on limits of HSM and so it is not known at what mass the effects begin or at which they are exacerbated. There is also limited data on the difference that small changes in mass can make.

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1.2 Injury considerations

The optimum position for any HSM is on the vertical (superior-inferior) axis connecting the COM of the head and the COM of the body. Any shift away from this axis increases the strain and fatigue on the neck muscles as they work to keep the head in an upright position [4]. The detriment of increased HSM on muscle fatigue and injury is greater when the COM of the HSM is further away from this vertical axis [13, 14]. Additional mass can lead to neck muscle fatigue which can reduce the protective efficiency of the musculature. It is also understood that repetitive loading can lead to intervertebral disc degeneration and that increased loads exacerbate this degeneration. However, these effects have not been quantified for small changes in HSM. The chronic neck injury risk associated with 3 kg and 5 kg HSM has been quantified using musculoskeletal modelling [13, 14]. Small reductions in the intervertebral stresses and repetitive injury risk were found for the 2 kg mass reduction when HSM has no anterior COM component. Hence, it is likely that these differences would be negligible when considering a small (i.e. 200 g) mass reduction. A large anterior COM component is expected when anteriorly-positioned helmet ancillaries such as night vision goggles (NVGs) are used. This study considers helmets alone, thus a small, or no, anterior COM component is of interest.

During traumatic injury events, such as falls or vehicle accidents, the addition of HSM can exacerbate neck injuries due to the increased MMOI. A limited number of studies have suggested that HSM may contribute to increased risk of neck injury in the case of an automotive frontal crash [15-17]. Consequently, minimising the total head supported mass may be an important strategy for minimising neck injury risk during a crash. However, although Merkle et al. showed that helmet mass was positively correlated with upper neck shear and tensile forces, they also showed that upper neck flexion, lower neck tension and lower neck shear were not significantly correlated to mass [17]. Doczy et al. showed that the effect of helmet mass was less important than impact deceleration [16] and Manoogian et al. suggested that an increase in mass, provided it was placed in the correct location, could actually reduce the neck injury risk [15]. However, if it was placed in the wrong location, it could increase the neck injury risk [15]. Merkle et al. utilised mass deltas of 900 g [17] and the mass deltas adopted by Manoogian et al. were not reported [15]. Although Doczy et al. adopted smaller mass deltas (227 to 460 g) the study was limited to sub-injurious loading due to the use of volunteer participants [16]. Thus, the influence of small HSM deltas on injury potential during frontal impacts remains unknown.

1.3 Aims

Technological advancements have resulted in small incremental reductions in helmet mass. Although reducing the mass of combat helmets has recognised benefits with regards to user performance and decreased risk of chronic and traumatic injuries, there is limited data quantifying the potential user benefits associated with small mass reductions. The aim of this research was to quantify changes in both user acceptance and traumatic injury risks associated with small (i.e. 200 g) changes in helmet mass.

2. METHODS

2.1 Human factors

Twenty Australian Army riflemen participated in the study. Three different helmet masses were employed. Each participant conducted one helmet condition on each trial day with the order of conditions balanced across the trial days. Each trial day comprised of the following activities:

- Two hours of representative military tasks:
    - 5 km pack march, into contact, drop pack
    - Section attack with blanks
    - Urban Assault with blanks
    - Obstacle course (comprising overs and unders, traverse ropes, monkey bars, balance walk, three tier tower, rope swing over, horizontal log walk, tunnel crawl, ten foot wall, leopard crawl, 'A' frame, burma bridge, cargo net crawl, balance cargo/net walk, and swinging bridge)
- Administration of questionnaire.

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A focus group was conducted at the end of the final trial day. During the trial days, participants wore combat uniform, boots, Australian body armour carrier with training soft armour and plates (plates were removed for the obstacle course as per standard operating procedures). Two gunners wore belt rigs in addition to the body armour system vest. Large field packs were worn during the pack march and cached prior to the other activities. Hearing protection was worn during the section attack and urban assault.

A carbon "bump" helmet had been modified to allow the addition of a mass adjustment rig (Figure 1). This rig, and specially designed helmet cover, allowed the mass of the helmet to be altered without participants being aware of the mass and with all other features of the helmet remaining the same. The mass was added centrally and vertically over the mid-point of the helmet, this would have the effect of slightly raising the COM of the system vertically but not affect the anterior-posterior or left-right positioning of the COM. Thus, the helmet mass was the main inertial variable to change across the conditions.

Figure 1. Trial helmets – mass adjustment rig and helmet cover

The base system was 1.0 kg, comprising the helmet, mass adjustment rig and helmet cover. Masses were added to produce the 1.2 kg and 1.4 kg conditions. Thus, the three helmet mass conditions were: 1.0 kg, 1.2 kg and 1.4 kg. Small system mass variations were expected due to variability in manufacturing of the helmet shell, mass adjustment rig and helmet covers. All test systems were weighed and found to be within 20 g of the target mass condition. The total HSM generally worn by the participants depended on the helmet size and which ancillaries were used. Based on these factors, the total HSM that the participants were accustomed to was likely between 1.2 kg and 2.2 kg, and thus encompassed the two heaviest test systems in this study.

A questionnaire was administered at the end of the two-hour trial. Six questions on discomfort, mass, mass distribution, difficulty in keeping head/chin up and overall user acceptance were addressed:

- The weight of the helmet was… (5-pt acceptability scale)
- The weight distribution of the helmet was… (5-pt acceptability scale)
- Did you experience discomfort to your head/neck whilst wearing the helmet? (Binary 'yes/no' response)
- The weight of the helmet made it difficult to keep my chin/head up when in prone… (5-pt agreement scale)
- Do you feel that the helmet slowed your head movement? (Binary 'yes/no' response)
- Overall, how acceptable was the helmet (as it is) to you as a user? (5-pt acceptability scale).

Given the low participant numbers, the alpha level was set at $\alpha = 0.10$ for all tests, where $p < 0.10$ was suggestive of a significant trend. Where questions used a 5-point acceptability scale or a 5-point agreement scale, non-parametric Friedman tests were conducted. Post-hoc analyses were conducted using Wilcoxon signed ranks when the Friedman tests indicated significance. The effect size was calculated using the Probability of Superiority, A. The A value is expressed as a percentage and indicates the probability that a person randomly chosen from Group 2 will have a higher score than a randomly chosen person from Group 1. Values of A equal to or greater than 56%, 64% and 71% were interpreted as small, medium and large effect sizes respectively [18].

Where questions used a binary 'yes/no' response, a Chi square test was conducted. The effect size for the binary response questions was calculated using the Odds Ratio (OR). The OR indicates the odds that someone with a given response came from a particular treatment group. OR values equal to or greater than 1.68, 3.47 and 6.71 were interpreted as small, medium and large effect sizes respectively [19].

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2.2 Neck injury risks while driving

2.2.1 Previous sled tests and models

The influence of additional helmet mass while driving was assessed using previously developed and validated MADYMO models of 50th percentile male and 5th percentile female Hybrid III anthropomorphic test devices (ATDs). The models were validated from a previous study comprising 13 sled tests with three Hybrid IIIs per test (two 50th males and one 5th female). The tests were conducted at four different impulses, varied HSM conditions (i.e. no helmet, two varieties of helmet and helmets with NVGs in different configurations) and with two different seats from Australian military vehicles. Initial testing showed significant rotation of the helmet on the head of the ATD during the impact, especially for the female ATD. To eliminate this motion and ensure a 'worst-case' rigidly coupled helmet mass, the helmets were subsequently bolted to the existing threads of the ATD heads.

A model of each of the seats was constructed in MADYMO. The surfaces of the models were captured with a coordinate measuring machine and were based on the geometry of the seats after they were installed on the sled but before they were used in any tests. To reduce the geometric complexity, only the centreline of the seats was modelled. The seats were then modelled with flat horizontal planes. The positions of the elements of the restraint systems were positioned in MADYMO at the locations recorded with the coordinate measuring machine. A finite element belt type was used where the belt and the ATD came in to contact. Line elements were used for the other parts of the belt that did not contact the ATD. The dimensions of the belts and their positions were also based on the measurements made with the coordinate measuring machine, while taking note of the belt routing around the ATD prior to the tests. The seat belt was modelled in two parts, a sash portion and a lap portion, tied at the buckle. The ATD was positioned in each seat to reflect the median position of the ATD across all tests (Figure 2). The body armour was modelled with multiple ellipsoids to represent the contoured front and back plates, with particular attention paid the areas where armour would come into contact with the belt or seat. The front and back plates were connected with a translational joint to model the straps. A helmet was modelled as a point mass with finite moments of inertia. This point mass was positioned at the location that represented the centre of gravity of the helmet. The point mass has several advantages in that it is simple to model, and that it represents the bolted helmet that was used in the sled tests. For the purposes of visualising the helmet in a model, an ellipsoid of no mass was added over the head. The MMOI of the helmet was based on that previously measured for a helmet of similar cut and was scaled to reflect the mass of the helmet under test.

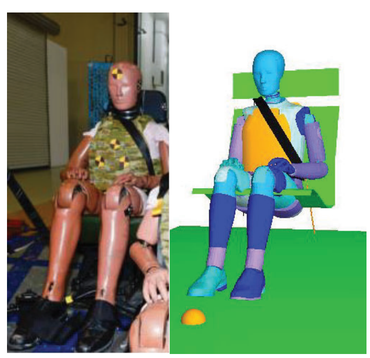

Figure 2. 50th percentile male ATD in seat with no helmet and corresponding MADYMO model

The models were adjusted to improve their concordance with the sled test result where no helmet was worn. The adjustments were limited to the gap between the seatbelt and the ATD, the length of seatbelt stored in the retractor, and the stiffness of the seatbelt. The length of seatbelt left in the retractor was adjusted until the payout of the seatbelt in the model closely resembled the payout of the seatbelt in the tests. Finally, the stiffness of the seatbelt was adjusted to provide a good fit to the apparent stiffness in the test results. It was found that this adjustment had a relatively small effect on ATD response. Three ATD kinematic variables were examined: the peak head acceleration, peak chest acceleration and peak pelvis acceleration; agreement between the sled tests and simulations in these variables indicated model validity.

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MADYMO simulations were completed to quantify the risk of neck injury with increasing impact velocities. Simulations were performed with each the 50th percentile male and 5th percentile female ATDs with the 1.2 kg helmet and 90 ms impact duration held constant and with varying impact speeds. These speeds were 20, 40, 50, 60, and 80 km/h. The results from these simulations were used to calculate regression equations for the neck injury metrics with varying impact speed (given a 1.2 kg helmet condition and 90 ms impact duration)(Table 1). Based on results of previous testing, the three most likely injury modes are lower neck (LN) flexion, upper neck (UN) flexion and the UN combined tension-flexion (N_{TF}); thus, these metrics are the focus of the current study.

Table 1. Previously developed linear regression equations describing neck injury metrics as a function of change in speed between 20 and 80 km/h, for 50th percentile male and 5th percentile female with the 1.2 kg helmet, with 90 ms impact duration

Regression coefficients	**LN Flexion (Nm)**		**UN Flexion (Nm)**		**UN N_{TF}**	
	Male	**Female**	**Male**	**Female**	**Male**	**Female**
a	6.76	6.20	1.61	2.09	0.012	0.03
b	83.16	30.48	62.07	5.80	0.04	-0.05
R^2	0.87	0.96	0.76	0.90	0.92	0.96

2.2.2 Current study

The impulse, helmet (1.2 kg condition) and seat used in the current study were included in the previous sled tests (13 male ATD and 6 female ATD runs) and subsequent MADYMO modelling. The seats were from tactical training vehicles which featured a standard 3-point harness comprising a lap and sash belt. To investigate the influence of small changes in helmet mass, five simulations were performed for each ATD with varying helmet mass and an additional test with no helmet worn. Helmet masses varied at 200 g intervals and comprised 0.6 kg, 0.8 kg, 1.0 kg, 1.2 kg and 1.4 kg. For the helmet tests, the COM was kept consistent for each test, with the value as previously determined for the 1.2 kg helmet. The MMOI was scaled according to the helmet mass**.** All simulations represented a 40 km/h impact with 90 ms duration 12.4 g deceleration pulse.

The neck injury metrics calculated were the UN tension, UN flexion, UN N_{TF}, LN tension, LN flexion and LN N_{TF}. Internationally-used thresholds exist for the male and female ATD for each metric, where exceeding the threshold generally represents a low (i.e. ≤ 5%) risk of a serious injury. Examples of serious injuries to the cervical spine include intervertebral disc rupture, major compression fracture of the vertebral body (no cord involvement) or spinal cord contusion with transient neurological symptoms.

3. RESULTS

3.1 Human factors

The human factors study results showed the following general tendencies: discomfort increased with mass, difficulty in keeping head/chin up when in prone increased with mass, acceptability of helmet weight decreased as mass increased, acceptability of helmet weight distribution decreased as mass increased, perceived speed of head movement decreased as mass increased and overall acceptability of helmet decreased as mass increased (Figure 3, Figure 4 and Figure 5). However, statistically significant differences were only reported for difficulty in keeping head/chin up when in prone, for the perceived weight of the helmet and for the perceived weight distribution of the helmet. Furthermore, whilst significant differences were reported between the 1.0 kg and 1.4 kg condition for these three parameters, only the difficulty in keeping head/chin up when in prone was significantly different at $\alpha = 0.10$ between 1.0 kg and 1.2 kg, and 1.0 kg and 1.4 kg. This demonstrates that a 400 g HSM difference had an effect on all three factors but 200 g only had a significant effect on difficulty in keeping head/chin up.

Only the 400 g mass difference resulted in medium or large observed effects, however small effects were observed between the 200 g conditions for most questions. The greatest effect size (large effect) was seen between the 1.0 kg and 1.4 kg conditions for the feeling of slowed head movement. Small effects were also observed between the 1.0 kg and 1.2 kg and the 1.0 kg and 1.4 kg conditions for experiencing discomfort, as well as between the 1.2 kg and 1.4 kg and the 1.0 kg and 1.4 kg conditions for overall acceptability.

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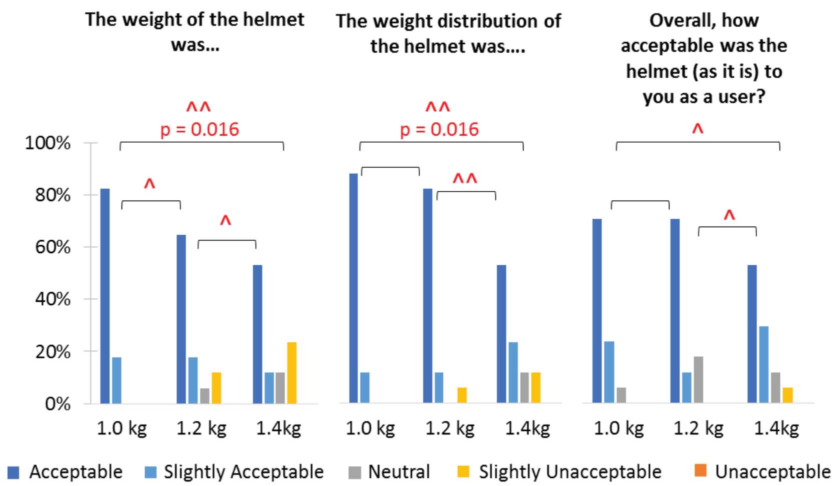

Figure 3. Results for questions answered using a 5-pt acceptability scale. Significance (p) only shown when p < 0.10, effect size, *A*, indicated if the requirements for small (^), medium (^^) or large (^^^) effects were met.

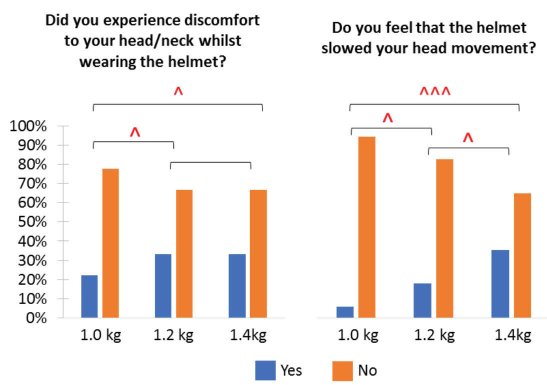

Figure 4. Results for questions answered using binary yes/no response. Significance (p) only shown when p < 0.10, effect size, OR, indicated if the requirements for small (^), medium (^^) or large (^^^) effects were met.

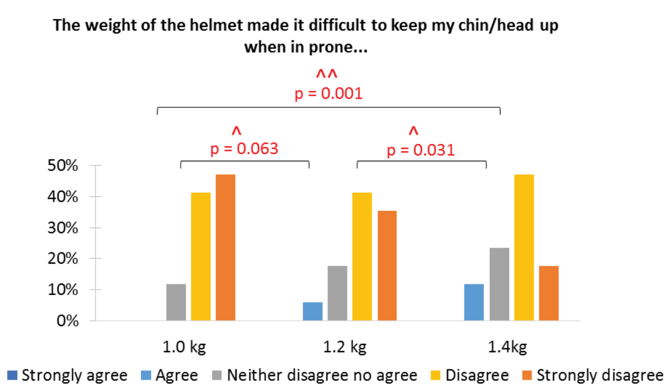

Figure 5. Results for the question answered using a 5-pt agreement scale. Significance (p) only shown when p < 0.10, effect size, *A*, indicated if the requirements for small (^), medium (^^) or large (^^^) effects were met.

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Regarding task performance, a small number of participants indicated that the helmet interfered with their ability to perform a pack march in the 1.2 kg and 1.4 kg condition but no statistically significant differences were identified. The qualitative comments indicated that the weight of the helmet made it difficult to keep the head up and there was a need to look down to obtain relief. The results of the rating scale data indicated that the weight of the helmets used in the trial did not appear to cause issues with performing a section attack, an urban assault or shooting. However, the qualitative comments indicated that some participants felt strain on their neck, particularly during pack marching, urban assault and section attack.

During the military tasks, the participants had been blinded to the helmet mass conditions. At the outset of the focus group, which took place at the end of the last trial day, participants were told that there had been three conditions. When asked what they thought the weight of the lightest helmet was, responses ranged from 300 g to 1 kg. When asked what they thought that the weight of the heaviest helmet was, responses ranged from 1.3 kg to 2 kg. When asked what they thought the weight difference was between each condition, there was general agreement that there was between a 50 g and 100 g difference.

3.2 Mass changes as equivalent speed reductions

The most likely injury modes were previously identified as LN flexion, UN flexion and the UN N_{TF}. Indeed for all simulations performed in the current test series, the only injury thresholds to be exceeded were these three injury metrics for the 5th percentile female and the LN flexion alone for the 50th percentile male. The LN flexion threshold is exceeded at all HSM conditions for the 5th percentile female and the no helmet condition is at 96% of the threshold (Figure 6). Comparatively, only the 1.2 and 1.4 kg conditions exceeded the LN flexion threshold for the 50th percentile male. The UN flexion and N_{TF} thresholds for the 5th percentile female were exceeded for HSM conditions of 1.0 kg and greater; the UN N_{TF} was at the threshold for 0.8 kg HSM (Figure 6).

For the 50th percentile male, the LN flexion threshold was exceeded for the 1.2 kg HSM condition but not the 1.0 kg condition (Figure 6). There were no other injury thresholds exceeded at the 1.2 kg HSM mass that was not exceeded at the 1.0 kg mass. Linear regression analysis was completed to predict the injury metric as a function of helmet mass for the given impact conditions (Table 2) as per Equation 1.

$$\textit{Injury metric} = a(\textit{helmet mass in kg}) + b \quad (1)$$

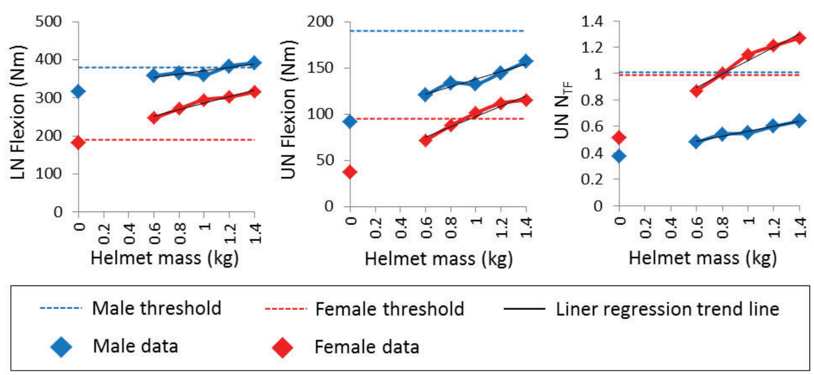

Figure 6. Influence of helmet mass on the injury metrics of LN flexion, UN flexion and UN N_{TF}. The 0 kg (no Helmet) data was not included in the linear regression due to the varied COM properties of the test compared to the helmeted tests. Note the 50th percentile male and 5th percentile female N_{TF} thresholds are both 1.0.

Table 2. Regression coefficients (a and b) and the coefficient of determination (R^2) for the three injury metrics of LN flexion, UN flexion and UN N_{TF} as a function of HSM (kg)

Regression coefficients	LN Flexion (Nm) Male	LN Flexion (Nm) Female	UN Flexion (Nm) Male	UN Flexion (Nm) Female	UN N_{TF} Male	UN N_{TF} Female
a	43.0	84.5	41.6	55.9	0.19	0.50
b	328.7	201.8	96.1	41.6	0.37	0.59
R^2	0.82	0.96	0.93	0.96	0.97	0.97

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Regression coefficient *a* indicates the change in the injury metric for each change in helmet mass of 1 kg. The LN flexion, UN flexion and UN N_{TF} data increases linearly with increasing HSM ($R^2 \geq 0.82$) for both the 50th percentile male and 5th percentile female. The regression results were used to calculate generalisable data regarding the influence of changing helmet mass.

Simultaneous solving of the equations of Table 1 and Table 2 permits the expression of the change or delta (Δ) helmet mass as an equivalent impact speed delta (Table 3). That is, for a 200 g reduction in helmet mass, the reduced risk of LN flexion injury for a midsized male was equivalent to that of travelling 1.3 km/h slower. Equally, with a 200 g lighter helmet, a midsized male could travel 1.3 km/h faster than otherwise with an equivalent risk of LN flexion injury. The change in HSM had a greater influence on the UN flexion and N_{TF} metrics, with a 200 g mass delta equivalent to speed deltas of 5.2 and 3.2 km/h respectively. Considering the LN flexion (i.e. the most likely injury mode) even a large mass change of 1.0 kg corresponded to a relatively small equivalent change in speed (6.4 km/h) for the same injury risk for a midsized male. Changing helmet mass had a slightly greater influence on neck injury risks for 5th percentile female than 50th percentile male. A helmet mass change of 200 g was equivalent to a change in speed of 2.7 to 5.3 km/h for the three injury metrics considered. Considering LN flexion, a large mass delta of 1.0 kg corresponded to a substantial equivalent speed delta of 26.7 km/h.

Table 3. The change in the injury metrics and equivalent changes in driving speed associated with helmet mass changes of 1 kg and 200 g for midsized male and small female occupants

	Δ Injury metric			Equivalent Δ speed (km/h)		
Helmet mass	**LN flexion (Nm)**	**UN Flexion (Nm)**	**UN N_{TF}**	**LN flexion**	**UN Flexion**	**UN N_{TF}**
Midsize male						
Δ 1kg	43.0	41.6	0.19	6.4	25.9	15.8
Δ 200g	8.6	8.3	0.04	1.3	5.2	3.2
Small female						
Δ 1kg	84.5	55.9	0.51	13.6	26.7	16.8
Δ 200g	16.9	11.2	0.10	2.7	5.3	3.4

4. DISCUSSION

This research aimed to quantify the changes in user acceptance and traumatic injury risks associated with small (i.e. 200 g) changes in helmet mass. Although other studies have investigated human factors and injury considerations for helmet masses which differ by larger amounts, decisions regarding helmet procurement are generally concerned with small differences between tendered items. Thus, these studies provide useful evidence with which helmets of plausibly different masses can be assessed and compared.

A 200 g change in helmet mass was found to have a significant effect on the difficulty of keeping the head up when in a prone position. No other significant differences were found due to 200 g mass changes and all effect sizes were small. A larger mass delta of 400 g caused significant differences in the acceptability of the helmet weight and weight distribution and the difficulty of keeping the head up with small to large effects found across all questions.

The study was limited to a sample of 20 participants and each helmet condition was only worn for 2 hours. It is likely that the small effects of the additional mass that were identified following 2 hours of activities would become larger, the longer the heavier helmet was worn. The results are, therefore, indicative and do not provide conclusive data on the implications of a 200 g HSM delta as worn by a large population or on an ongoing basis. As per the ergonomic recommendations in Ivancevic and Beagley, an HSM with the smallest mass is preferable [1]. The human factors study addressed the basic mass of the helmet; however the helmet also functions as a platform for mounting ancillaries such as NVGs, counterweights, torches, and cameras, which would substantially increase the total HSM. At higher masses and varied head-HSM system COMs, a reduction in mass of 200 g would potentially yield greater human factors benefits.

HSM reductions of 200 g were found to reduce the risk of injury equivalent to quite small reductions in driving speed of 5.3 km/h or less for both midsized males and small females. For midsized males, the only injury threshold exceeded was LN flexion at the 1.2 and 1.4 kg mass conditions. The changing HSM had slightly greater effect on the 5th percentile female injury metrics;

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the UN flexion and UN N_{TF} thresholds were exceeded for helmet masses 1 kg and greater and the LN flexion threshold was exceeded for all helmet conditions assessed. However, the no helmet condition represented 96% of the threshold value, indicating injury risk for small females at the tested impact condition even without HSM. The aforementioned injury predictions are limited to the impact conditions modelled in the simulations. This impulse was derived from publically available crashworthiness data of the Ford F150 truck. The speed equivalencies for changing helmet mass are limited to driving speeds of 20 to 80 km/h and impacts with a deceleration duration of 90 ms. The deceleration duration is affected by the stiffness and mass of the vehicle, whereby vehicles with greater stiffness during impacts, or lighter mass, have shorter impact durations.

Frontal vehicle impacts cause forward flexion of the neck. Therefore, a COM that is more anterior and superior can exacerbate the flexion and tension forces. However, helmets alone can result in the COM moving posterior to the natural COM of the head. Thus, equivalent rearward vehicle impacts may be the more dangerous impact scenario for this HSM condition. Further, the neck is less tolerant to extension than flexion bending moments. However, rearward vehicles impacts are generally at lower impact speeds than frontal impacts.

The injury analyses assessed the neck injury risk with helmets alone and in a specific scenario. If the helmet was used as part of a system including NVGs, it would have the effect of moving the COM anteriorly. The anterior translation of the head-helmet COM would likely exacerbate the chronic injury risks [13, 14], traumatic injury risks during frontal impacts and potentially the traumatic injury risks during events with lower peak decelerations e.g. falls.

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5. CONCLUSIONS

Human factors data indicates that 200 g reductions in HSM resulted in small improvements in the human factors areas considered in this report. The traumatic neck injury risk while driving was slightly reduced with 200 g HSM reductions, equivalent to driving 5 km/h slower or less. This report addressed one type of traumatic injury and participants only worn the helmets for 2 hrs. The results and conclusions of this report are, therefore, bounded by this scope. Small mass differences may have greater impacts during other traumatic or chronic injury modes and on other human factors considerations. Further, a 200 g HSM delta may have a greater effect when the effect of wearing the helmet is an anterior shift of the head-HSM COM or when the total system mass is greater. From both a human factors and injury perspective, any reduction in mass is seen as beneficial.

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