Optimization of Muscle Activation Schemes in a Finite Element Neck Model Simulating Volunteer Frontal Impact Scenarios

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Keywords: Muscle activation, human body model, impact biomechanics, cervical spine, optimization

Word count: 2903

Abstract

Neck muscle activation is increasingly important for accurate prediction of occupant response in automotive impact scenarios and occupant excursion resulting from active safety systems such as autonomous emergency braking. Muscle activation and optimization in frontal impact scenarios using computational Human Body Models have not been investigated over the broad range of accelerations relevant to these events. This study optimized the muscle activation of a contemporary finite element model of the human head and neck for human volunteer experiments over a range of frontal impact severities (2g to 15g). The neck muscles were grouped as flexors and extensors, and optimization was undertaken for each group based on muscle activation level and activation time. The boundaries for optimization were defined using data from the literature and a preliminary parametric study. A linear polynomial method was used to optimize the model head kinematics to the volunteer experiments for each impact severity. The model predicted muscle activation to increase with higher impact severities, and the resulting optimization improved the average cross-correlation by 35% (0.561 to 0.755), relative to the original model activation scheme. Importantly, a newly proposed cocontraction activation scheme for maintaining the head in a neutral posture provided a 23% improvement in correlation compared to the original model. This study identified an activation scheme to obtain more accurate response kinematics in computational Human Body Models as well as contributing to the understanding of muscle influence during impact scenarios.

1. Introduction

Muscle activation plays a role in injury risk of the cervical spine, acting to stiffen the spinal column during the application of internal and external forces (Stemper and Corner, 2016). In addition, the neck musculature surrounding the spinal column enables the dynamic movement and stability of the head and neck. Yet the understanding of the muscle activation schemes in the context of spinal injury risk during motor vehicle accidents requires further development (Olszko et al., 2018). This is important information as approximately one-quarter of the four million emergency room visits per year from motor vehicle traffic injuries in the United States are associated with neck and back strains and sprains (Albert and McCaig, 2015). Further, one study conducted in Sweden showed that frontal car impacts are related to one-third of all cervical spine injuries (Kullgren et al., 2000).

Clinical studies reviewed by Stemper and Corner (2016) examining spinal injuries following car crashes have found that pre-awareness of the human occupant before impact reduces neck injury risk, symptoms, the intensity of the pain and recovery time (Hendriks et al., 2005; Sturzenegger et al., 2012); although this evidence is mixed, with others reporting no correlation (Walton et al., 2013). Vehicle safety tests commonly rely on anthropometric test dummies or post mortem human subjects (PMHS) unable to represent the true muscle activation behavior (Arbogast et al., 2009; Iwamoto, 2018; Stemper and Corner, 2016). However, some level of muscle activation must be present during real-life crash scenarios, which could alter the kinematic response of the occupant. The influence of this muscle activation contributes to the limited extrapolation of vehicle safety tests on how they would compare to those of human volunteers (Albert et al., 2018).

Recently, the influence of the muscle activation during autonomous braking and frontal impacts were investigated using a low severity impact sled or car tests with volunteers (Ólafsdóttir et al. 2013; Fanta et al. 2013; Kumar, Narayan, and Amell 2003; Carlsson et al. 2010; Siegmund et al. 2007; Hedenstierna, Halldin, and Siegmund 2009; Dehner et al. 2013; Mathews et al. 2013). A widely referenced study consisting of 16 volunteers subjected to 119 frontal and 72 lateral impact sled tests (Wismans et al., 1986) was conducted by the Naval Biodynamics Laboratory (NBDL). While the NBDL data set comprises a range of impact severities, no muscle activation was recorded from the participants during the tests. The tests reported only the head and neck kinematic response of human volunteers ("National Highway Traffic Safety Administration," 2012).

Computational Human Body Models (HBM) using finite elements (FE) offers the potential to examine load sharing and injury risk of different anatomical structures during a variety of loading scenarios without the ethical and practical challenges of using living human subjects (Dibb et al., 2013). Several HBM have been used to investigate neck muscle activation, and there is an increasing interest in including a closed-loop control for these muscles (Östh et al., 2015), which could enable a model to adapt for different loading conditions. However, one of the challenges in this process is the scarcity of electromyography data in high severity impacts (>8G) and, therefore, many HBM are calibrated to specific tests and loading regimes. Some researchers have optimized the cervical muscle activation for different scenarios (Dibb et al., 2013; Ivancic and Pradhan, 2017; Mortensen et al., 2018), but no attempt to observe the effects over a range of low to high impact severities has yet been considered.

Many HBM of the neck have previously implemented some form of muscle activation scheme, such as the KTH model, JAMA model, THUMS model, and simTK model (Brolin et al., 2005; Ejima et al., 2005; Iwamoto et al., 2012; Vasavada et al., 1998). Except for the THUMS model, these implementations did not incorporate a combination of skin, adipose tissue, 3D passive muscles and active 1D muscles. Further, all of these studies, including the THUMS model, have focused only on lower severity events when validating against volunteer studies.

The M50-O v4.5, 50th percentile, Global Human Body Models Consortium (GHBMC) HBM incorporates skin and adipose tissue as well as two types of elements to represent the muscles. These elements are 1D Hill-type elements and 3D Ogden rubber elements. The first element represents the active properties of the muscle, and the last element represents the passive properties. They are connected to the hard tissues and each other in the model with support elements so the muscle can follow physiological lines of action.

The model presented a default activation for the 1D active muscles and, in this study, it was named Maximum Muscle Activation (MMA) (Shen and Cronin, 2017). It corresponded to a neuronal impulse (step signal) of 100% for 100ms generating a maximum value of the activation function in the muscle of 0.871 (Panzer et al., 2011). The onset time for this activation was defined by the average (74ms) of the values presented in the literature for a series of tests with volunteers. In the present study, the MMA level was 100% and the activation levels of other activation schemes were expressed as a percent of the MMA level.

This study aimed to optimize the muscle activation parameters in a detailed HBM based on the NBDL frontal impact data for the head and neck kinematics using open-loop control of the neck musculature activation levels over a range of impact severities. It was hypothesized that optimal muscle activation schemes for different impact severities could be identified based on correlation with the experimental volunteer data and that, due to the higher accelerations, a unique activation scheme could present good correlation to all the severities.

2. Methodology

A previously developed and validated model of the head and neck was extracted from an average stature male HBM (Figure 1) (M50-O v4.5, 50th percentile, GHBMC). The neck model included detailed representations of the neck musculature, represented by passive 3D and active 1D Hill-type elements, the cervical vertebrae (C1 to C7), the first thoracic vertebra (T1), cervical spine ligaments, intervertebral discs, and skin. Since the current study was focused on neck response, a representative model of the head with a rigid skull was integrated with the neck. This model had the same mass (4.4 kg) and center of gravity as the original detailed head model but provided reduced computation time necessary for the optimization undertaken in the current study.

The linear forward X-acceleration and Y-rotational displacement boundary conditions from the volunteer experiments were applied to the first thoracic vertebra (T1) in the model. The NBDL experimental head kinematics were reported for individual cases and in this study were averaged for comparison to the model output. The average curve and standard deviation for each impact severity (Figure2) were generated using point-wise analysis for each case within a specific impact severity.

Nine different frontal impact severities were simulated, ranging from 2g to 15g (Ewing and Thomas, 1972; Thunnissen et al., 1995), for a total duration of 250ms. The neck muscles were activated in two separate groups: flexors and extensors, following a strategy implemented in previous work on active neck musculature (Panzer et al., 2011) (Table 1).

The 1D muscle elements generate a contraction force that is dependent on the physiological cross-sectional area (PCSA) of the muscle, material properties related to the force-length and force-velocity responses, the muscle activation level related to the neuronal signal received by the muscle and the muscle activation time. As the PCSA and material properties were already defined in the model based on previous studies (Bogduk and Yoganandan, 2001; Pettersson et al., 1997; Yoganandan et al., 2002), the current analysis focused on the muscle activation level and time. The overall shape of the activation level versus time curve was based on the literature (Happee, 1994). The muscle activation onset time (Figure 3) was varied from 55 to 99ms, representing the range of values presented in the literature (Blouin et al., 2003; Foust et al., 1973; Hernández et al., 2006; Magnusson et al., 1998; Ono et al., 1997; Siegmund et al., 2002; Snyder et al., 1975). The muscle group activation level was varied between 0 (no activation) and 100% (the existing activation strategy) (Figure 4).

A series of initial simulations were undertaken to determine the sensitivity of the FE model to the level of activation and establish a range of boundary conditions for the optimization study. First, the flexors activation level was varied from 0% to 100%, maintaining the extensors with constant low activation of 10%. It was identified that the model response was not sensitive to the flexor activation level for impact severities above

3g, attributed to only the extensors being antagonistic to the head impact response and the lower strength of the flexors to the extensors (Figure 5).

The extensors activation level was varied from 0% to 100%, maintaining the flexors with 100% activation, as recommended in earlier studies (Panzer et al., 2011) to identify ranges of activation for the optimization study (Table 2). From these initial simulations, three activation schemes were defined: MMA, No Muscle Activation (NMA) (i.e. lower bound), and a Cocontraction Muscle Activation (CMA) scheme; where the muscle activation was set to maintain the head in a neutral posture without any external influence. The CMA scheme comprised a 100% flexor activation and an extensor activation level, using the same activation curve shapes presented in Figures 3 and 4, which maintained the head in the neutral position when no external loading was imposed on the model.

In the optimization study, a linear polynomial optimization method with D-optimal point selection and domain reduction was applied using commercial optimization software (LS-OPT, R.5.2.1, LSTC, Livermore, CA) to identify an activation scheme for each impact severity (Dibb et al., 2013; Ivancic and Pradhan, 2017). The simulations were undertaken with a commercial FE solver (LS-DYNA, LSTC, R.7.1.2, Livermore, CA). The activation time range (55 to 99ms) was based on reported values for the sternocleidomastoid and trapezius. The variation for the extensor group activation scaling was based on the results from the initial parametric study (Table 2).

The model was optimized to the head CG kinematics (X-linear displacement, Y-rotational displacement, X-acceleration and Y-rotational acceleration) as reported in the human volunteer tests using the mean square root method. The optimization was considered converged when the average of the mean square root values varied less than

1%. The correlation between the model head kinematics and the experimental data was determined using cross-correlation as implemented in a widely used software program (CORA, Partnership for Dummy Technology and Biomechanics, R. 3.6.1, Germany).

3. Results

The head and neck model was assessed for all nine impact severities and compared to the human volunteer data, with an average cross-correlation rating of 0.561 using the MMA scheme (Figure 6) for all impact severities.

The initial parametric study provided approximate activation values for each impact severity and served to identify the boundary values used in the optimization study. Following this study, the CMA strategy was investigated. The CMA was a single activation strategy, a subset of the cases considered, where the head remained in a neutral position when the muscles were activated with no external loading applied. This activation scheme comprised flexors activated at 100%, extensors activated at 21%, with an onset time of 74ms producing an overall correlation of 0.691 (Figure 6).

The NMA presented an overall correlation of 0.651, close to the CMA, mainly due to the high correlation rating for low severity impact scenarios (e.g. 2g and 3g impact severities).

The optimization study was undertaken for the model at each impact severity to identify the required extensor activation level resulting in the highest correlation for each impact severity (Figure 6). In general, the optimized activation level increased with higher impact severities (Figure 7).

The correlation of the head center of mass kinematics using the OMA scheme (Table 3) with the volunteer data was higher than all other schemes for all severities tested, as expected Figure 6. All OMA onset times occurred at the lower boundary for impact severities at or below 8g, indicating an earlier onset time was desired. However, the lower boundary was maintained at 55ms, which was the lowest published value for measured activation time in humans. There was no observable trend for the OMA times for the higher severity impact cases, and the onset times ranged from 64 to 81ms. This variation was attributed to the lower sensitivity of the high severity cases to onset time.

Discussion

The rising interest to develop active safety systems in the automotive industry has increased the demand for improved understanding of the physiological response of the human body under a range of scenarios, from low severity sudden stops to high severity frontal impacts. The current study investigated muscle activation levels for human volunteer tests over a range of frontal impact severities using a FE head and neck model from a contemporary HBM. Moreover, this was the first study to optimize muscle activation levels and onset times for a broad range of frontal impact severities.

The optimization method presented in this study for the muscle activation in frontal impact improved the correlation of the kinematic response of the model with the experimental data for all severities on average by 35% (0.561 to 0.755, respectively fair and good biofidelity according to ISO/TR9790) reaching a maximum improvement of 103% for the optimized 3g impact case. The early onset times of the optimized responses relative to the MMA are likely related to muscle tonus before the impact in the experimental cases that were not present in the current GHBMC neck model formulation.

The electromyography percentage of the maximum voluntary contraction of the trapezius in low severity experiments (1g) was 10% (Hedenstierna et al., 2009), similar to the OMA for the extensors in the 2g and 3g impacts. The fixed activation during this study presented an activation of 100% which is much higher than the 40% normalized electromyography level presented in the literature (Hedenstierna et al., 2009), but the flexor muscles did not strongly affect the head kinematics for the higher severity impacts (>3g). For these lower severities impacts, the NMA scheme presented a similar correlation to the optimized case, which may indicate that other soft tissues present in the model, such as the skin and flesh, are too stiff and produce increased resistance to the forward movement of the head, effectively decreasing the required activation levels. Future research will continue to investigate the effect of soft tissue on low severity impact response.

The model was also run with the NMA scheme resulting in an overall correlation of 0.651. Although this correlation was higher than the MMA, the stress levels identified in the ligaments and vertebra indicate the potential for tissue failure for the higher severity cases (>12g), which was not observed in the volunteer tests, and therefore this was not considered an improved overall correlation.

The OMA schemes were, on average, just 6% better correlated to the kinematic data than the CMA, which may indicate that two mechanisms are acting in the neck during the crash: an almost constant response (CMA) with higher contribution to the kinematics and an activation related to the magnitude of the head acceleration with lower contribution to the kinematics. These inferences are in agreement with the literature (Happee et al., 2017), which shows that there are two main mechanisms related to neck stabilization, the vestibulocollic and cervicocollic reflex. However, the contribution of these two mechanisms are almost the same in that study, which could be attributed to the slower accelerations compared to the ones from impacts that were used. The optimized scheme could help guide future research on activation strategies for impact scenarios while providing an improved kinematic response for the current model.

Another contribution of this study is the deeper comprehension of the parameters affecting the impact response and how they should be handled during the development of closed-loop or open-loop motor control for muscle activation. For example, the results indicate that the flexors have little effect on the neck kinematic response in high severity frontal impacts, which was expected as this muscle group cannot generate forces counteracting the forward motion of the head.

The improved overall correlation rating with the experimental data, of the CMA scheme agrees with the initial hypothesis that a single activation scheme, during impacts, could be used to obtain a head kinematic response with good correlation to the experimental data.

Future studies should consider the potential impact due to subject variability as the current study only optimized activation to the average subject response. Another limitation of this study was that the muscles were divided into only two groups, and each group had all the bulk elements activated in the same manner. This was an oversimplification of the problem; muscles in the same group do not necessarily activate similarly and muscles may not act exclusively in one group during neck movements. However, as the improved correlation score shows, the hypothesis that the response of the muscle groups is dominated by the strongest muscles (sternocleidomastoid and trapezius) is reasonable for global kinematic studies.

4. Conclusion

In conclusion, the findings of this study provide new insight into the role of muscle activation on the impact response of the head and neck during frontal impacts. This data provides a recommended trajectory-based activation scheme depending on impact severity and could be used to further elucidate the influence of muscle activation and onset time of the flexors and extensors in head kinematics in other impact scenarios, especially for the lower severities in which the muscle influence is higher.

Acknowledgments

This study was made possible due to the financial support provided by the Natural Sciences and Engineering Research Council of Canada, Global Human Body Models Consortium, Honda R&D Americas, FCA Canada and GM Canada; the head-and-neck model provided by the Global Human Body Models Consortium; and the computing resources provided by Compute Canada.

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Figure captions

Figure 1: 50^{th} percentile male neck model with active extensors (blue) and flexors (red) including the SAE coordinate system utilized in this study.

Figure 2: Example of experimental average and standard deviation for head center of gravity Y-rotational displacement for the 8g frontal impact.

Figure 3: Different activation schemes for different neuronal input levels (NI) with different onset times (55ms, 74ms and 99ms) for the extensor muscle group.

Figure 4: Different activation levels for the flexors with a constant activation level of 10% for the extensors in the 8g frontal impact.

Figure 5: Preliminary study activation schemes and Optimized Muscle Activation correlation rating with the experimental data for each impact severity.

Figure 6: X-linear displacement and Y-rotational displacement in relation to T1 for the activation schemes for 2g, 8g and 15g impact severity.

Figure 7: Optimized extensors activation level for all simulated severities.

Figure 1

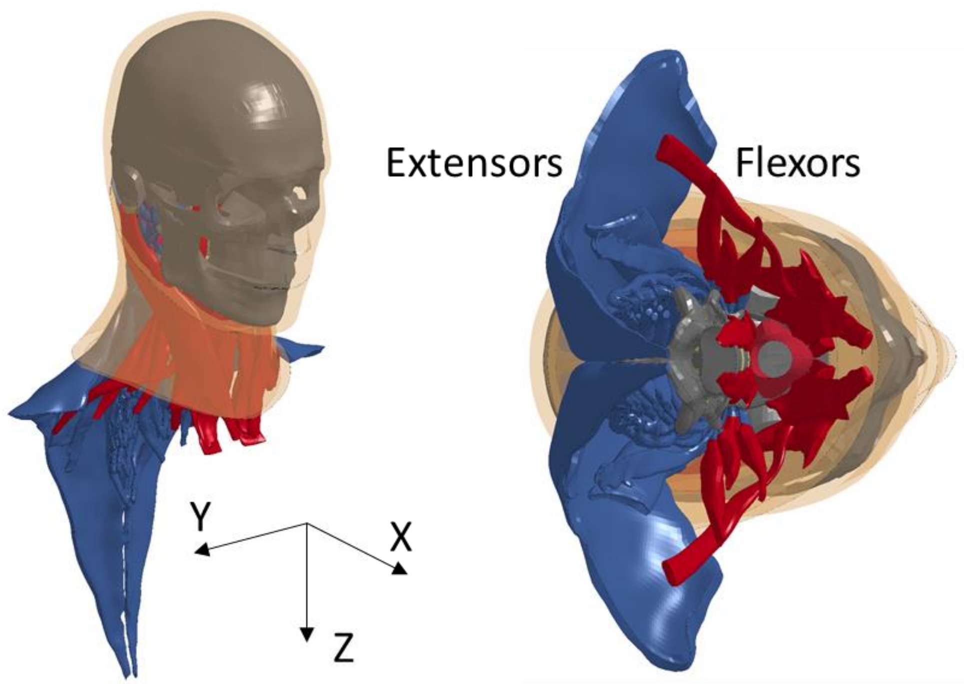
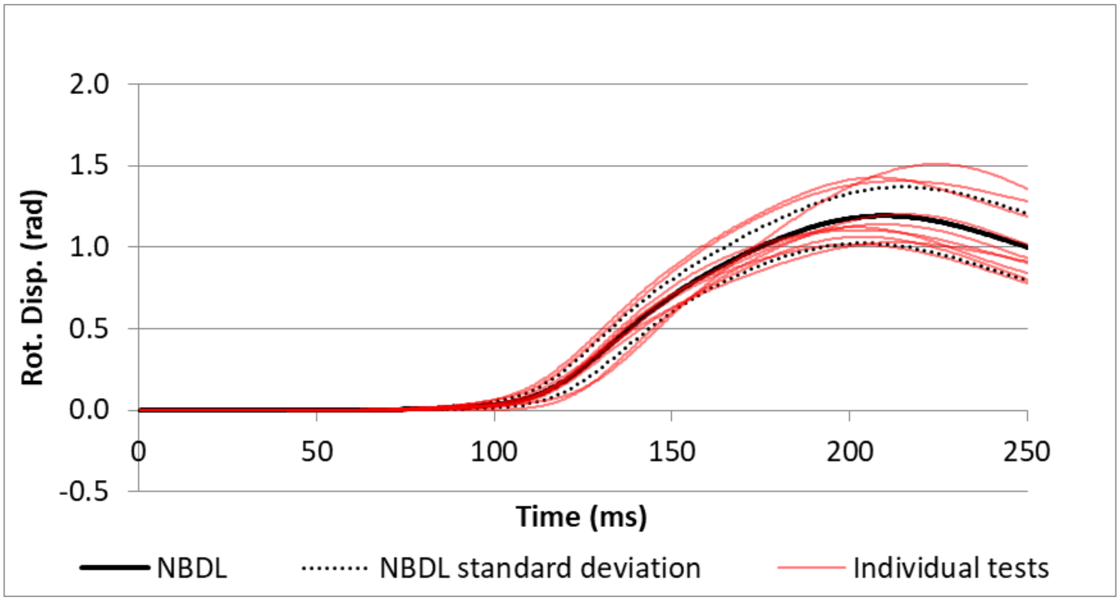


Figure 2


2.0
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200
250
Time (ms)
NBDL
NBDL standard deviation
Individual tests

Figure 3

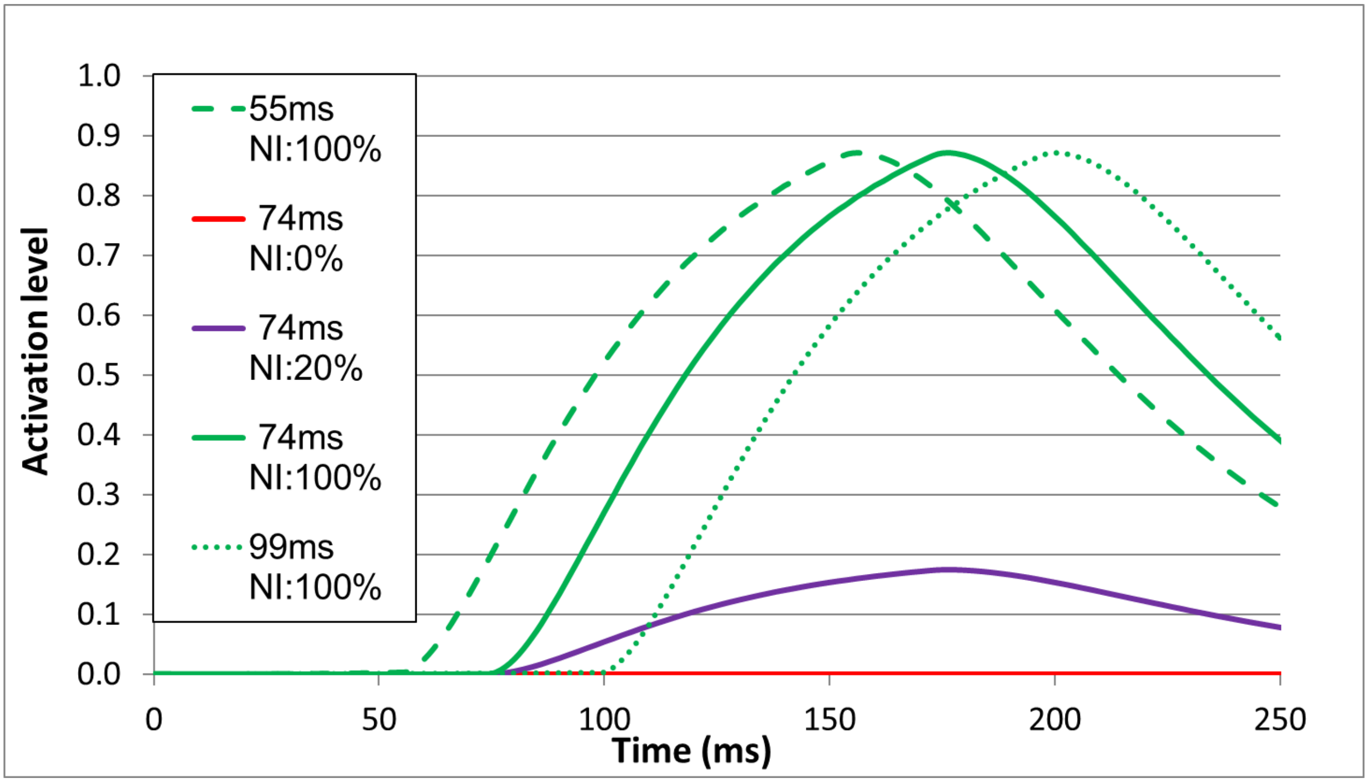

1.0
0.9
0.8
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0.0
Activation level
55ms NI:100%
74ms NI:0%
74ms NI:20%
74ms NI:100%
99ms NI:100%
0
50
100
150
200
250
Time (ms)

Figure 4

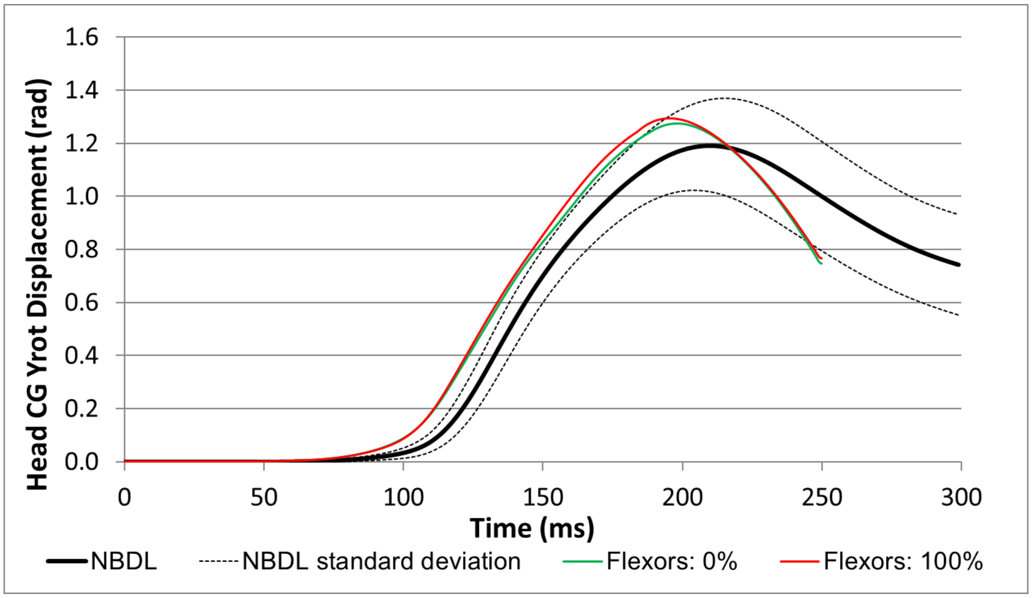

1.6
1.4
1.2
1.0
0.8
0.6
0.4
0.2
0.0
Head CG Yrot Displacement (rad)
0
50
100
150
200
250
300
Time (ms)
NBDL
NBDL standard deviation
Flexors: 0%
Flexors: 100%

Figure 5

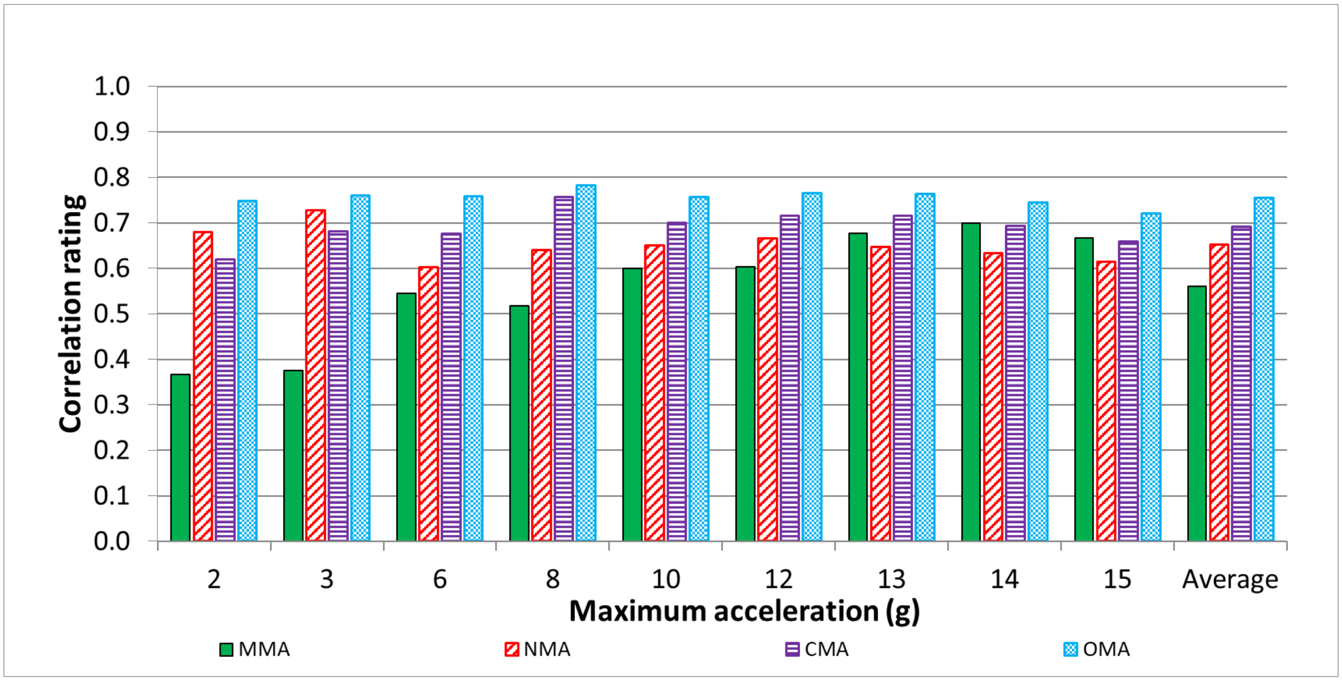
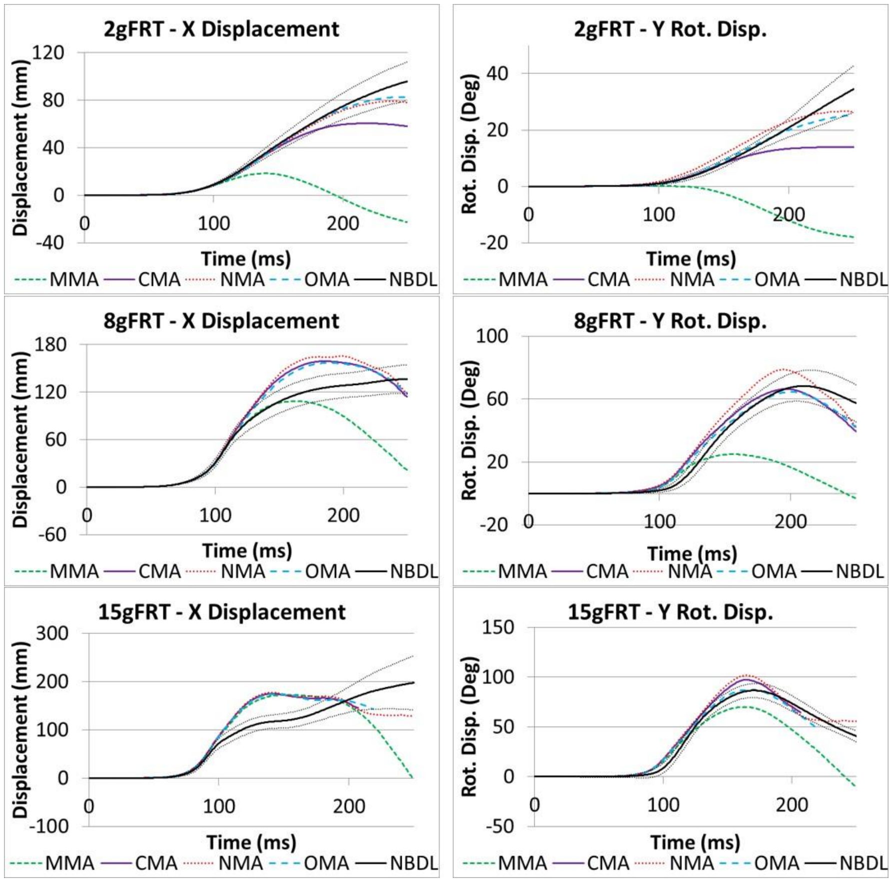

1.0
0.9
0.8
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0.0
Correlation rating
2
3
6
8
10
12
13
14
15
Average
Maximum acceleration (g)
MMA
NMA
CMA
OMA

Figure 6



2gFRT - X Displacement
Displacement (mm)
Time (ms)
2gFRT - Y Rot. Disp.
Rot. Disp. (Deg)
8gFRT - X Displacement
8gFRT - Y Rot. Disp.
15gFRT - X Displacement
15gFRT - Y Rot. Disp.
MMA CMA NMA OMA NBDL

Figure 7

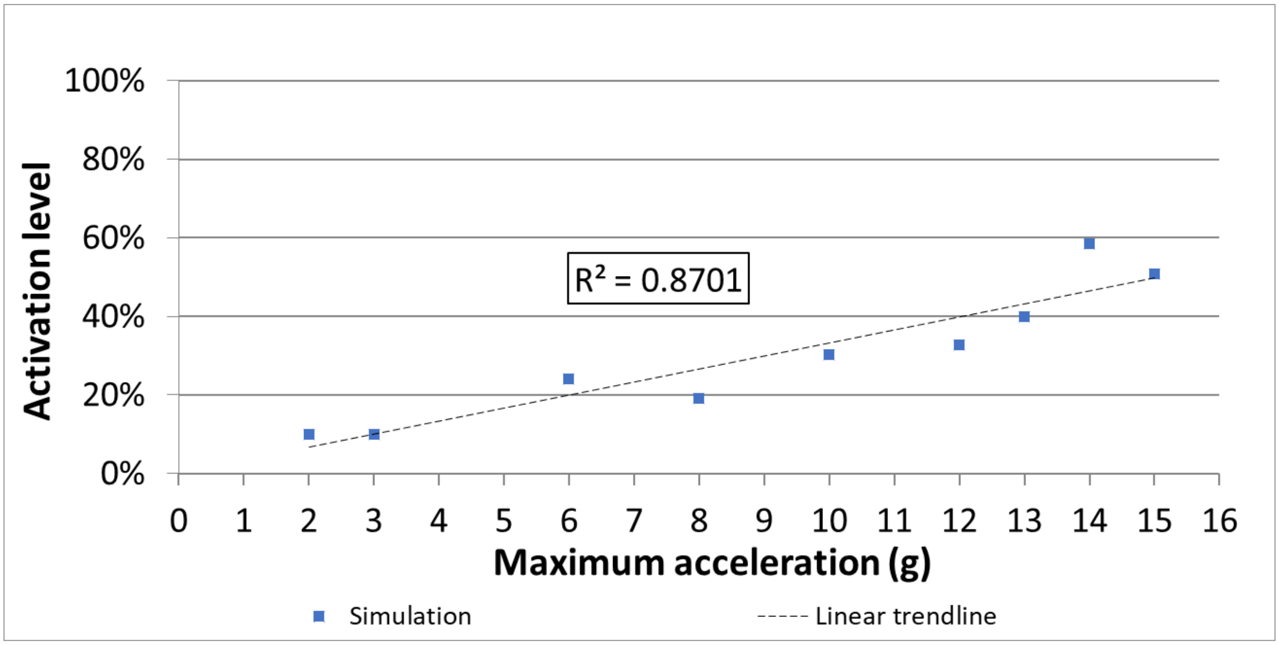

100%
80%
60%
40%
20%
0%
Activation level
R² = 0.8701
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16
Maximum acceleration (g)
Simulation
Linear trendline

Table 1: Muscle group definitions in the neck model.

Muscle group	
Extensors	Flexors
Oblique capitis inferior	*Longus capitis*
Oblique capitis superior	*Longus colli*
Rectus captis major	*Rectus capitis anterior*
Rectus captis minor	*Rectus capitis lateral*
Iliocostalis cervicis	*Anterior scalene*
Longissimus capitis	*Middle scalene*
Longissimus cervicis	*Posterior scalene*
Multifidus	*Sternocleidomastoid*
Semisplenius capitis	*Omohyoid*
Semisplenius cervicis	*Sternohyoid*
Splenius capitis	
Splenius cervicis	
Levator scapulae	
Rhomboid minor	
Trapezius	

Table 2: Initial parametric study results for approximate muscle activation levels versus impact severity.

Impact severity (g)	Number of individual experiments	Extensor activation scaling	Flexor activation scaling
2	10	0.0	
3	17	0.0	
6	18	0.2	
8	10	0.2	
10	19	0.4	1.0
12	10	0.4	
13	10	0.4	
14	10	0.6	
15	8	0.6	

Table 3: Muscle parameters of each activation scheme.

Impact severity (g)	MMA	NMA	CMA	OMA		
				Flexors activation scaling	Extensors activation scaling	Extensors onset time (ms)
2	Flexor: 1.0 Extensor: 1.0 Onset: 74ms	Flexor: 0.0 Extensor: 0.0	Flexor: 1.0 Extensor: 0.21 Onset: 74ms	1.0	0.10	55.00
3					0.10	55.00
6					0.24	55.00
8					0.19	55.00
10					0.30	78.38
12					0.33	68.63
13					0.40	64.04
14					0.58	81.10
15					0.51	77.85