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A Literature Review of Musculoskeletal Injuries to the Human Neck and the Effects of Head-Supported Mass Worn by Soldiers

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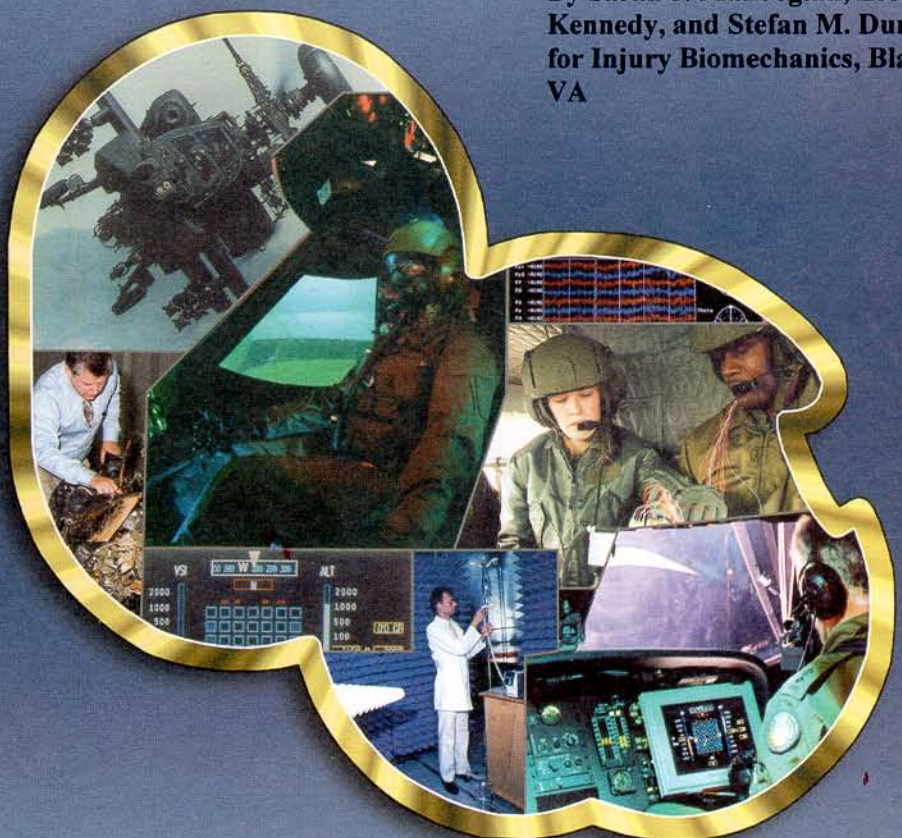

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14. ABSTRACT

This report reviews research, conducted prior to 2004, in four areas of neck biomechanics that pertain to head-supported mass (HSM). First, computer models of the head and neck are reviewed for their biofidelity and versatility. Second, neck injury criteria are reviewed for their applicability to studying HSM-related neck injuries. Third, existing studies of HSM effects on head-neck response to impact are reviewed. Fourth, a review of muscle activation and its effects on neck impact response is presented. This study found that HSM poses an undermined risk to the safety of the Soldier. Computational simulations exist to accurately simulate human response to a variety of loading conditions, including simulations of live occupant response with passive or active muscles. Neck injury criteria can be used to determine the susceptibility of the simulated occupant to potential neck injury. Further studies utilizing these tools will ensure that military personal are not subjected to unnecessary safety risks as a consequence of additional head-supported mass.

15. SUBJECT TERMS

Head supported mass, neck injury, cervical spine injury, aviation, aircrew helmets, NVGs

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1. Introduction

Cervical spine injuries can be incurred under a variety of circumstances. In the past 30 years, particular emphasis has been placed on reducing the number and severity of cervical spine injuries received by automobile accident victims and by military aviators. Over the same time span, technological advances in military equipment have resulted in more devices being mounted on the helmet to enhance the capability of the soldier. They include night vision goggles (NVGs), counterbalance weights, chemical masks, oxygen masks, information display visors, communications equipment, and more. As these systems are mounted to helmets, the soldier's neck must support this head-supported mass (HSM) and the resulting dynamic characteristics of the head and neck system are changed. It is widely accepted that these systems increase the likelihood of both low and high-severity injury, but the additional risk of neck injury that these systems create has not been quantified.

As the understanding of neck injuries becomes more advanced, efforts must be taken to quantify the physiological risk from adding HSM to soldiers. These efforts should come to reduce the risk of injury and increase effectiveness on the battlefield. Barazanji et al. (1998) summed up the present situation best when he stated, "Safe and tolerable limits of head-supported device mass properties, such as mass location and distribution are important design criteria for future aircrew helmets. The obvious challenge for the Army research community is to establish those safe limits for HSD mass properties that can be tolerated by male and female aviators alike."

This report presents a literature review to identify the latest developments in four main areas of neck biomechanics relative to HSM injuries. First, mature computer models of the head and neck were reviewed to determine the most biofidelic and versatile model available. Second, neck injury criteria were reviewed and analyzed for applicability to the study of HSM related neck injuries. Third, existing studies that analyzed the effects of head-supported mass related to head-neck performance in impact scenarios were reviewed. Finally, a review of muscle activation and its effects on neck impact response was conducted. It should be noted that this literature review focuses on existing adult neck work, and child neck studies are not included in this report.

2. Mature Computational Models of the Head and Neck

2.1. Introduction to Computational Modeling

As computers became practical for studying complicated mathematical problems, researchers have employed computers to aid in the understanding of the physical phenomenon they are studying. Recent advances in computer technology have allowed the use of complex computer simulation tools in biomechanics research and offer an invaluable alternative to experiments. Once a computational model has been validated, it can be run repeatedly, allowing a detailed study of the effects of minor changes to the total system performance. It can generate data useful to the researcher for the development of injury criteria and the determination of injury risk under specific conditions. Computational models are useless, however, unless it is proven that they can accurately replicate results from experimental testing. There are three main types of computational simulations used today: multibody models, finite element models, and combination models where multibody and finite element models are used in conjunction for computational efficiency.

2.2. Multibody Models

Multibody models are the simplest of the three computational models. In multibody simulations, rigid bodies are connected to develop a complete model of the biomechanical mechanism or structure of interest. Each of these rigid bodies may have different inertial or stiffness properties and can exert forces on adjacent rigid bodies.

Huston et al. (1978a, 1978b) developed a 3-D computational model of the head and neck for use in whiplash-type injury studies (Table 1). At the time it was the most sophisticated head-neck model available. This model uses a series of rigid bodies to represent vertebral bone, and springs and dampers to represent passive muscles, ligaments and cervical discs, resulting in a 54 degree of freedom head-neck model. This model can be used to simulate the head-neck system response to a simulated impact.

Table 1 - Computational Multibody Models

Author (Year)	Program	Loading Condition	Advantages	Limitations
Huston (1978)	Unknown	Frontal impact; Lateral impact; Rear impact	Accounts for passive muscle; Good correlation with limited validation (PMHS & volunteer)	Lumped parameters for discs/ ligaments/ muscles; Not a full body model; Needs further validation; No active muscles; No injury prediction
Deng (1987)	DYNCOMBS	Frontal impact; Lateral impact; Rear impact	Passive muscles with separate lines of action; Good correlation with flexion and lateral volunteer data	No active muscles; Not a full body model; No PMHS validation; Muscle updates required; No injury prediction
Williams (1983)	Unknown	Frontal impact; Lateral impact	Separate elements for discs/ ligaments/ muscles; Active muscles; Good correlation with frontal and lateral volunteer data	Not a full body model; No injury prediction; No PMHS validation
Jakobsson (1994)	MADYMO	N/A	Rear impact	Computationally efficient; Sufficient validation for qualitative assessment of occupant response
Bomar (1998)	Head-Spine Model (PC)	Frontal impact; Lateral impact; Rear impact; Vertical impact	Separate elements for discs/ ligaments/ muscles; Graphical user interface	Inaccurate material properties; Not validated; Not a full body model; Currently in development

Huston et al.'s (1978a, 1978b) model was validated against two experimental groups; one with Post Mortem Human Subjects (PMHS) and one with live volunteers. In practice, this model achieved "excellent agreement" between model simulations and the limited experimental data that was available. Parameters used for validation were: head angular acceleration, angular velocity and angular displacement. Considering the state of technology at the time, the model is very good, but its usefulness is limited, as it was created before well founded injury criteria had been determined.

Deng and Goldsmith (1987) developed a 3D lumped parameter model of the head-neck and upper torso for use in the DYNCOMBS software package (Table 1). Their model utilizes the Huston et al. (1978b) approach for solving the relative motion between bodies, Lagrange's form of d'Alembert's principle (Deng, 1987). This model also uses passive muscle pairs. The muscles are massless, and in order to approximate muscle curvature in the neck, the muscles use 3-point lines of action. It was compared to volunteer frontal and lateral flexion experimental results and correlates well. It was speculated that most of the differences between the computational model and the experimental results can be explained from the simplified computational muscle modeling and the fact that volunteer responses include active muscle contraction. In order to gain a more biofidelic model, Deng and Goldsmith (1987) stated that "improvement of muscle modeling incorporating the proper mass and geometry should be a most crucial goal in future investigations."

Williams and Belytschko (1983) created another 3D rigid body model of the cervical spine (Table 1). Rigid bodies of vertebrae are connected by deformable elements representing the discs, facet joints, ligaments and muscles. Unlike Huston et al.'s (1978a, 1978b) model, Williams and Belytschko created their model with curved musculature which could be left to operate passively or set to respond actively after a certain time of 40 to 100 ms into the simulation. Overall, the simulations show that active muscle behavior can have a great influence on model behavior compared to passive muscles only. This model was validated for frontal and lateral impacts and good correlation with volunteer experimental data validate the active muscle behavior modeling of the computer simulation.

Jakobsson et al. (1994) developed a MADYMO model for use in studying occupant performance in rear-end collisions (Table 3). However, this model is a greatly simplified two dimensional rigid body model. As the item of interest was a study of “whiplash” type trauma, the spine is modeled by a series of rigid bodies. The model was validated against a limited series of rear-end volunteer simulations and determined to be biofidelic enough for a qualitative assessment of occupant response in rear-end impact scenarios. The forces best correlated to risk of injury were tensile and shear forces between vertebrae, head angular acceleration, and volume rate of change of the cervical spinal canal (Jakobsson, 1994).

The most recent multibody model developed for neck injuries is the U.S. Air Force Head Spine Model by Bomar and Pancratz (1998) (Table 1). The model is an update of an existing Air Force Head Spine Model, so that it may be run on a personal computer platform. The model consists of rigid inertial elements with massless deformable elements to represent muscles, ligaments, cartilaginous joints and other connective tissues. According to Bomar and Pancratz (1998), data for some of the element properties of the original Head Spine Model are several orders of magnitude different than currently available material property data. These discrepancies were not corrected during this particular revision, as the focus was on programming the existing Head Spine Model for use on a modern computer platform. In addition to corrections of element properties, recommended updates by the creators include more realistic muscles and invertebral-discs as well as more accurate assumptions of the static tension in cervical ligaments. Future updates will allow the addition of rigid bodies to the existing model and will include injury estimate processing capabilities (Bomar, 1998).

The Bomar and Pancratz Head Spine Model is still in development, but it promises to have good potential for head-spine work. The model discussed has not been validated, to date, and does not utilize the latest published material properties, nor does it have the capability to allow additions of other rigid bodies to an existing model, for analyses such as head-supported mass studies. Other multibody models have been developed for studying specific impact loading conditions on the neck; however, they do not have significantly different features than the models already discussed (McElhaney, 1979; Goldsmith, 1984; Tien, 1985; Paver, 1990; Bowman, 1984; Seemann, 1984; Bowman, 1981; Melvin, 1972; Bowman, 1972; Li, 1991; Bosio, 1986; Bowman, 1975; Tien, 1987; Nightingale, 2000).

2.3. Finite Element Models

Finite Element (FE) models are more complicated than the multibody models and require more computational time, but can provide more detailed information than multibody models. In FE modeling, the geometry of a body is defined and broken down into a discrete grid of elements. Stresses and strains can be calculated within elements and localized regions of stress and strain can be determined. Forces or accelerations can then be exerted on the model, to pinpoint the areas where stress and strain may be high. The disadvantage of this type of modeling approach is that it is not as computationally efficient as multibody modeling and thus simulations cannot be run as quickly or in as many configurations as easily as with a multibody model.

LS-DYNA was used by Kleinberger (1993) to develop a 3D model of the cervical spine (Table 2). To increase its accuracy, this model needs additional musculature, improvements to soft tissue material properties and refined geometry. Run times are up to 20 hours in duration per simulation for a cervical spine model only. The model was given only a limited validation against a set of published experimental data. It was stated that testing was underway to provide experimental data for further development and validation, but more recent publications discussing this model are not presently available.

Table 2 - FE Model Summary

Author (Year)	Program	Number of Elements	Loading Condition	Advantages	Limitations
Kleinberger (1993)	LS-DYNA	1600 Solid/ Vertebrae	Frontal impact; Lateral impact; Rear impact	Uses published material properties; Detailed mesh geometry	No musculature; 20 hour runtime; Not validated; Not a full body model; No injury prediction
Dauvilliers (1994)	RADIOSS	150 Solid; 104 Shell; 412 Damping-spring	Frontal impact; Lateral impact	Good correlation with volunteer data on initial impact; Computationally efficient; Integrated with FE full body model; Global injury prediction in full body model	No musculature; Poor correlation with volunteer data after initial impact; No PMHS validation; Full body model intended to represent 60 year old male
Nitsche (1996)	PAM-CRASH	1852 Solid; 86 Membrane	Frontal impact; Lateral impact; Axial impact	Good correlation with PMHS and volunteer data	No musculature; Not a full body model; No injury prediction
Yang (1998)	PAM-CRASH	11,498 Solid; 3071 Shell-membrane	Frontal impact; Rear impact; Axial impact	Detailed geometry; Passive muscles included	More validation required; Not a full body model; 3-24 hour runtime
Deng (1999/2002)	LS-DYNA	Unknown	Frontal impact	Detailed geometry; Active muscles; Good correlation with low-severity frontal impact	Not a full body model; More validation required; Enhancement of material models needed
Halldin (2000)	LS-DYNA	4560 Solid; 3572 Shell; 230 Spring	Axial impact	Able to predict injury from compressive impact	No musculature; Transverse processes not included on vertebral bodies; Not a full body model; Limited loading conditions; 45 hour runtime

Another FE model of the human neck was developed by Dauvilliers et al. (1994) using RADIOSS software (Table 2). The model was developed for use in frontal and lateral impacts. The developers modified ligament stiffness in an attempt to include passive neck muscles in the model, as they were thought to affect dynamic behavior of the head and neck (1998). The FE neck was then integrated into a 50th percentile seated male finite-element model by Lizee et al. (1998). The goal was to have a full body FE model of a seated male. However, the model needs more development to predict injury risks (Lizee 1998). Dauvilliers et al. (1994) also noted that more realistic passive muscle action is needed in future versions of the model, for a more biofidelic response.

Nitsche et al. (1996) developed a FE model of the spine utilizing PAM-CRASH software (Table 2). This model is of the cervical spine only, and it does not contain active or passive muscles. Such a basic model, however, illustrates a problem with current FE models of the head and neck region. Since local parameters such as max stress and strain are unknown for the cadaver and volunteer validation tests, the model cannot be validated against its calculated output. The model was validated against global motion of the neck with the assumption that if the global motion is correctly simulated, the local stresses and strains of local tissue must be reasonably accurate, and are therefore validated (Nitsche, 1996).

Nitsche's (1996) model was validated by comparing to published experimental data of frontal flexion, lateral flexion, and compression of the spine in both volunteer and cadaver experiments. Experiments of both frontal and lateral flexion were compared to the PAM-CRASH model's displacement of the occipital condyles and the center of gravity of the head relative to a non-rotating T1, in order to determine the relative rotation angle of the head. Comparison of the simulation's output to the experiments show that the FE model motions displays acceptable agreement with test results in frontal flexion, but less agreement in lateral flexion (Nitsche, 1996). In order to gain more realistic biofidelity and usefulness, this model needs to be enhanced with musculature and integrated into a head-upper torso system.

The most recent 3D FE model of the neck was developed by Yang et al. (1998) using PAM-CRASH software (Table 2). This model is a full FE version of the head and neck, which can be incorporated into an upper torso model. The intended application of this model is to study the neck loads experienced as an occupant comes in contact with an airbag. This model includes passive muscle modeling only, with no active muscle response. This model is very computationally intensive as it requires 3-24 hours on a Cray supercomputer to run one simulation. For initial validation this model was been compared to a limited amount of cadaver tests with encouraging results; however, this model needs more validation against experimental data before it can be fully utilized. Also, reductions in run times or advances in computer technology need to be made to maximize this model's usefulness.

Deng et al. (1999) developed a LS-DYNA FE model of the neck with detailed 3D anatomical data (Table 2). This model contains detailed intervertebral discs with both the nucleus pulposus and the annulus fibrosis, neck ligaments, and detailed 3D representations of the vertebral bodies. All material properties used for the model are based on numerical analysis of existing published data. Although the vertebral bodies are modeled with elastic-plastic material properties, all validation work on the model has treated the vertebral bodies as rigid materials. An update of this model added neck musculature to the FE model (Deng 2002). The muscles are modeled as Hill-type elements and allow for active muscle generation. The updated model with muscles was validated against published low-severity frontal crash volunteer studies. The model displays good general agreement with the volunteer tests once the muscle activation schemes have been optimized.

A 3D FE model of the neck was developed by Halldin et al. (2000) to study compression injuries of the neck (Table 2). This model runs in the LS-DYNA software environment and utilizes 4560 solid elements, 3572 shell elements and 230 spring elements to model the cervical vertebrae, ligaments and discs. Since neck musculature is not thought to greatly influence neck response on compressive impacts, neck musculature and the corresponding transverse processes of the vertebral bodies are omitted from this model. This model was validated to axial impacts of the head and was found to be able to predict injury by local stress of neck tissues; however, at certain impact angles the model predicted failures where none were experienced in the corresponding experimental test. This model has not been validated for any other type of impact situation, nor have the authors mentioned intent to further develop the model for other loading scenarios.

Other finite element models have been developed to study specific impact loading conditions on the neck. However, they do not have better features than the models discussed previously (Roychoudhury, 2000; Yang, 1992; Choi, 2002).

2.4. Multibody-Finite Element Combination Models

Perhaps the most effective method to study a particular region on a large scale model, multibody-finite element combination models allow the user to model the area of interest with finite element techniques and other global regions with computationally efficient multibodies. Several software packages allow this option, with LS-DYNA and MADYMO being among the most commonly used.

Table 3 - Multibody-Finite Element Combination Models

Author (Year)	Program	Number of Elements	Loading Conditions	Advantages	Disadvantages
De Jager (1996)	MADYMO	N/A	Frontal impact; Lateral impact	Active muscles; Detailed geometry; Good correlation with frontal and lateral volunteer and PMHS data	Unrealistic muscle lines of action; Less sophisticated than Van der Horst model
Camacho (1997)	LS-DYNA	639 Rigid; 448 Deformable	Axial impact	Good correlation with axial loading cadaver data; Computationally efficient	No musculature; 3-24 hour runtime; Not fully validated; Not a full body model
Van Ee (2000)/ Chancey (2003)	LS-DYNA	639 Rigid; 448 Deformable	Axial impact	Active muscles; Good correlation with axial loading cadaver data	Not fully validated; Not a full body model
Van der Horst (2002)	MADYMO	N/A	Rear impact; Frontal impact; Lateral impact	Good correlation with frontal, lateral and rear impact volunteer and PMHS data; Refined geometry over De Jager model; Active muscles; Detailed for local and global injury assessment	FE techniques would offer better local injury assessment; Lack of high severity muscle activation data

Camacho et al. (1997) created an LS-DYNA model of the cervical spine and head (Table 3). The intended use of this model is to simulate spinal behavior for “near-vertex” (+/-15 degrees of head vertex with torso) head impacts (i.e. compressive forces). To this effect, the attached head is modeled as a deformable finite element head with rigid body vertebra. Like many of its computational predecessors, this model has no neck musculature. According to the authors, this is because “under the conditions of near-vertex impact, injuries occur two to three times more quickly than the muscles of the cervical spine react” (Camacho, 1997). The authors also assert that due to the lack of data available on dynamic material properties, many of the material properties had to be inferred. It is for this reason that it makes sense to use a more computationally efficient lumped or rigid body model, that does not rely on complete material properties to determine accurate three dimensional kinematics. Also, they state that material-based tolerance criteria lack injury correlation with predicted stresses and strains.

The Camacho model was updated by Van Ee et al. (2000) to include neck musculature (Table 3). This update added 24 muscle pairs to the model via spring elements. The muscle response characteristics are based upon the physiologic cross-sectional area of the muscles in the cervical spine. Basing the muscle response characteristics on the size of the muscle and incorporating detailed lines-of-action of the individual muscles is thought to provide the most physiologically accurate estimation of each muscle’s contribution to the motion of the cervical spine. The muscles were also modeled such that they could be actively controlled in the simulation. The model was only validated to tensile neck testing and the active effect of the muscles was shown to move the site of injury typically seen at the lower cervical spine in experimental cadaver work, to the upper region of the cervical spine where most clinical cases of spine injury are observed. This model was again updated by Chancey et al. (2003) to determine the state of active musculature required to maintain the head in an initially stable, upright position. A common problem with active musculature modeling is that the head is not initially stabilized, due to the effects of gravity, prior to the crash event. This model was used to more accurately estimate the tensile neck tolerance of the cervical spine, though it has not been validated to other loading scenarios.

The most current detailed neck model, created in MADYMO, was completed by Van der Horst (2002). The Van der Horst (2002) model is a major update of the MADYMO head-neck model developed by de Jager (1996). The de Jager model traces its heritage back to the previously discussed three-dimensional head-neck multibody model of Deng and Goldsmith (1987). In order to develop a detailed head and neck model for the MADYMO software package, de Jager adapted Deng and Goldsmith's 3D head and neck model (Table 3). De Jager preferred to use a more simplistic model than a FE model, since FE models are very complex, computationally inefficient and difficult to validate with so many parameters. Therefore, the model was implemented as a discrete parameter model with multiple rigid bodies, from which the complexity could be increased as the model was validated. De Jager intended that a computationally efficient, validated model, the end result would be more practically useful than an FE model, since material properties of the human neck are not fully known (de Jager, 1994).

The de Jager model offers fair agreement for frontal impact, its inaccuracies attributed to inappropriate modeling of AOC joint. Reasonable to excellent agreement with experimental results are found with lateral impacts (de Jager, 1994; de Jager, 1996a). Results from de Jager's model show that head rotation in this model is too large, most likely as a result of inadequately stiff muscles. Comparing to PMHS experiments, the "cadavers show a similar difference in response with the volunteers as the model, indicating that muscle tensioning limits head rotation and prevents overtipping for the volunteers" (de Jager, 1994).

De Jager's (1996b) detailed neck model also includes the capability to simulate active muscle control. Fourteen pairs of Hill-type elements are used as muscles in the detailed model. De Jager also developed a global model, which has lumped parameters for discs, ligaments, facet joints and muscles. While certainly more detailed than the global model, the muscles in the detailed model are still not completely representative of anatomical geometry. The muscles in the detailed model are not curved, and are attached only to an average vertebrae position (de Jager, 1996a). The inclusion of neck musculature means that modifications to neck strength can be made by changing parameters characterizing the muscles. This may be very useful for studying the effects of HSM on soldiers who strengthen neck muscles versus soldiers that do not, for example. Also, due to the increased anatomical description of the neck in the detailed model, the user has the ability to determine loads and deformation of individual soft tissue within the neck (de Jager, 1996a). Several "next steps" were identified by de Jager for future improvements for an even more refined model. These recommendations include a more detailed intervertebral joint, separate representations of soft tissue, and more refined geometry, including an increase in the detail of the neck musculature (de Jager, 1994). Attempts to improve the de Jager model were carried out by Yamazaki et al. (2000) and Brelin-Fornari (1998); however, the updates were not as significant as a more recent update.

A major update of the De Jager model was performed by Van der Horst (2002) (Table 3). The Van der Horst model, like the De Jager model, includes rigid bodies as vertebrae but includes more detailed geometry of facet joints and ligaments, as well as anatomically accurate, curved neck muscles. Although the MADYMO software program is capable of finite element modeling, the Van der Horst model only uses multi-body techniques (Van der Horst, 2002). Van der Horst's model uses Hill-based muscles, which are currently the most widely used and accepted mathematical adaptation of a muscle (Van der Horst, 2002; Winters, 1990a). In de Jager's neck model, the neck muscles are modeled by cord elements connecting the muscle attachment points. Van der Horst added significantly more detail to the neck muscles as the cord elements do not accurately simulate active neck muscles (Van der Horst, 2002). Multi-segment muscles were added to allow for curvature of neck muscles and therefore more realistic lines of muscle action (Happee 1999; Van der Horst, 2002).

Van der Horst's neck model has been incorporated into a larger, full body human model (Happee, 1999; Van den Kroonenberg, 1997). The intent was to create a biofidelic MADYMO human body model for use in a variety of omnidirectional computer simulations. Once validated, experiments can be conducted using either dummies or cadavers and correlated to the computer model of the human subject using active muscles. A validated computer model that could easily be modified would be useful to study aspects like body size, posture, muscular activity and post fracture response (Happee, 1999), or the effects of HSM. Also, the computer model anatomy is detailed enough, despite the lack of FE techniques, that it can give insight into injury mechanisms on a tissue level (Happee 1999). In 2000, Happee et al. published a paper on two recent MADYMO models, a small female and mid size male. The Van der Horst detailed head-neck model is used only in the mid size male model. However, the male model has been widely validated using frontal volunteer sled tests, frontal and lateral PMHS impactor tests, lateral PMHS sled tests, and rearward volunteer and PMHS tests (Happee, 2000).

The Van der Horst model was used for extensive testing in rear-end impact simulations and gave very encouraging results (Van der Horst, 2001) (Table 4). Data shows more realistic responses with stiff passive muscles than with normal passive muscles, except for head center of gravity x-displacement which shows too little x-displacement. Therefore, it is logical that for future work, the stiff muscles be used to obtain a more accurate biofidelic solution (Van der Horst, 2001).

Table 4 – Biofidelity of Stiff Passive Muscle Response in MADYMO Detailed Human Neck Validation Tests (Van der Horst, 2001)

Good (in envelope)	Reasonable (<25% outside envelope)	Poor (>25% inside envelope)
Head rotation (stiff muscles)	T1 Rotation	T1 Z-displacement
Head CG x-displacement (normal muscles)	T1 X-displacement	Head CG acceleration (>100ms)
Head CG z-displacement (stiff muscles)	Head rotation (normal muscles)	
Head CG z-acceleration	Head CG x-displacement (stiff muscles)	
Head CG angular acceleration	Head CG z-displacement (normal muscles)	
	Head CG acceleration (initial)	

Van der Horst's model was also validated against volunteer data for frontal crashes from 2 g to 15 g. Muscle contraction shows a large influence on the head-neck response. Wismans et al. (1998) describes high severity frontal crash simulations with PMHS. In testing, it is noted that head center of gravity trajectories are of the same order of magnitude as lower severity volunteer experiments, but the head rotations are larger in the PMHS. This is attributed to the fact that the PMHS do not benefit from the active muscle control that volunteers in low severity testing are able to display (Wismans, 1998). Similarly, as suggested by the volunteer and PMHS experiment comparisons, it is noted in testing of Van der Horst's model that muscle contractions have a large effect on the head-neck response (Wismans, 1998).

In a 15 g frontal simulated crash volunteer test, it is found that the Van der Horst model with active muscles predicts accurate head-neck response in terms of trajectories, head rotation and head lag. In the same testing, it is noted that angular and resultant head acceleration are largely unaffected by active muscle response (Wismans, 1998). For this reason, the authors conclude that acceleration data may not be a good indicator of true biofidelity in a model. This may also mean that injury criterion such as the Neck Injury Criterion (NIC), discussed later in Chapter 3, will not be affected by active muscle response, since it is based in large part on head acceleration.

In lower severity simulations, in order to achieve realistic performance of the simulation compared to the volunteer experiments, a lower activation level of the muscles and a larger reflex delay may be required (Wismans, 1998). Similarly, the influence of the muscles on the occupant simulation increases with the muscle activation level (Van der Horst, 1997). By this reasoning, it can be speculated that the effect of muscles during severe impacts is not yet fully understood. Since volunteer testing is limited to low severity testing, muscle activation in high severity impacts has not been fully tested. Computer simulations may offer the greatest insight into this phenomenon if data can accurately be extrapolated for full muscle activation.

Wismans et al. (1998) concluded that the Van der Horst model with fully activated muscles lies almost entirely within response corridors developed by volunteer experiments for frontal impact scenarios. This computer model was also determined to be more biofidelic than both the THOR and Hybrid III Anthropomorphic Test Devices (ATDs). Further validation of the Van der Horst model was conducted by Van Hoof (2002) and was found to correlate well with volunteer test data. The muscle activation levels in the Van der Horst model will likely have to be optimized for studies involving HSM, but it is expected that the impulse is likely to be relatively high severity and that 100% muscle activation will yield the most accurate response.

Lateral validation tests were performed for the Van der Horst model by Meijer et al. (2003). In these tests, volunteer responses to low severity lateral impacts were recorded and head kinematics were used to validate the simulation output. By varying the levels of muscle activation to between 50 percent and 100 percent activation the model displayed good correlation to the head kinematics measured by the volunteers. These validation experiments were performed with only two volunteers, so further validation work is planned for lateral impacts, although initial results have shown to be promising.

Seating posture was also varied by Van der Horst for evaluation of the effect of occupant positioning on neck response. From the outcome of the studies, it was shown that initial posture has a large influence on the head-neck motions (Van der Horst, 2001; Van der Horst, 2002). Therefore, for studies involving HSM, it is imperative that proper positioning of the occupant, such as the aviator, be considered before running the simulation. Also, likely out-of-position scenarios should be evaluated to determine if they represent a significant risk of injury beyond the normal "in-position" crash sequence.

Other multibody-finite element combination models have been developed for studying specific impact loading conditions on the neck; however, they do not have features other than those already discussed (Hayamizu, 1999; Weerappuli, 1998).

2.5. Computational Model Summary

Computer models have been used to better understand the human head-neck system kinematics since the 1970's. Although favorable correlation was found even then, computer models of today display a great deal of biofidelity and offer versatility and repeatability not possible experimentally.

Finite element models are capable of providing detailed information regarding localized loading conditions and predicting high stress or strain areas. FE models are also very computationally intensive, requiring longer amounts of computing time per simulation.

Multibody models, however, are more computationally efficient than FE models, and offer biofidelic responses on par with current FE element models. Additionally, multibody-finite element combination programs, such as MADYMO, allow users the versatility to choose between computational efficiency and generation of detailed localized analysis. They allow the user the option to develop a multibody model, which can be modified later if material properties are not known, or if more detailed output is desired.

Van der Horst (2002) has developed the most widely validated head-neck model using the MADYMO computational simulation program. This model has the capability of simulating active muscle response, which sets it apart from other models available today. Due to the lack of available information on all material and failure properties of the neck, developers of FE models are forced to iterate material property values until they reach dynamic correlation with volunteer and PMHS experiments. Therefore, the Van der Horst detailed neck model loses very little in terms of output, but offers a great deal in its validation and active muscle features. It is currently the model of choice to use in a computational simulation of neck response to dynamic loading conditions.

3. Neck Injury Criteria

3.1. Injury Criterion

As knowledge of injury mechanisms and maximum physiological limits of human occupants is gained; it is to be expected that several different injury criteria to be developed, depending on factors such as injury mechanism, the acceleration environment, or impact condition to which the occupant is subjected. Further, establishing tolerance levels for exposure to these injury mechanisms is complicated, given the variations in tolerance from person to person and the variation in tolerance among different loading mechanisms (Patrick, 1987). For neck injuries in particular, there are several injury criteria, for the prediction of both serious and minor injuries. Singular loading criteria and combined loading criteria (N_{ij} and Modified N_{ij}) are used to predict the likelihood of serious neck injuries according to the Abbreviated Injury Scale Score. This scoring system defines a serious neck injury as AIS3+. Criteria for minor extension or flexion of the neck (NIC, IV-NIC, and N_{km}) are used to predict the likelihood of minor (AIS1) neck injuries.

3.2. Singular Loading

Some types of impact provide a direct uniaxial force or single plane bending of the neck. For instance, a swimmer diving into a pool may impact their head directly on the pool floor, forcing the neck into compression (McElhaney, 1979). For predominantly unidirectional loading of the neck (tension, compression, shear) there are maximum limits established for neck injuries. Also, for bending of the neck in the sagittal plane, there are maximum values for flexion and extension bending moments. These values were determined largely through the work of researchers such as Mertz et al. (1971) who performed testing on human volunteers and cadavers to determine physiological limits of the body to such loading conditions. These tests were conducted both statically and dynamically to provide the most accurate and useful data for impact loading conditions. For the purpose of automobile safety testing, these values were then converted to representative values that are applicable to crash test ATDs. Maximum values for these loading conditions for a 50^{th} percentile male ATD are detailed in the Federal Motor Vehicle Safety Standards (FMVSS) (Digges, 1998) (Table 5).

Table 5 - Allowable Neck Loading in FMVSS 208 as Measured in the 50% Male ATD (Digges, 1998)

Loading	Allowable
Axial Compression (N)	4000
Axial Tension (N)	3300
Fore & Aft Shear (N)	3100
Flexion Bending Moment (N-m)	190
Extension Bending Moment (N-m)	57

As a result of recent feedback to the National Highway Traffic Safety Administration, new limits have been placed on axial compression and axial tension loads (Eppinger, 2000). These were originally proposed to place a limit on maximum tension and compression for combined loading analysis, but are applicable only for combined loading where little or no bending loads are present (Table 6).

Table 6 - Current Maximum Axial Loads as Measured in the ATD (Eppinger, 2000)

Symbol	Loading	95% Male ATD*	50% Male ATD	5% Female ATD (In-Position)	5% Female ATD (Out-of-Position)
F_{max}	Peak Tension (N)	5030	4170	2620	2070
F_{max}	Peak Compression (N)	4830	4000	2520	2520

***** 95% Male ATD is not included in final ruling, but performance limits were given for informational purposes

A study performed by Nightingale et al. (1997) examined dynamic spinal behavior when subjected to buckling under compressive impact loading conditions. Testing was performed on 22 head and neck cadaver specimens and both flexion and extension attitudes were invoked. After initial buckling of the spine, various bending modes were induced. The authors theorize that the variety of induced bending modes may explain why compressive head-neck injuries occur at different vertebral levels and with widely varying mechanisms. It was also noted that the compressive strength for male neck cadaver specimens that exhibit buckling under compressive loading was significantly higher than for females, 2243±572 N versus 1061±273 N respectively. In another study, Pintar et al. (1995a) found 3800 N to be the failure force under pure compressive loading for males.

Alem et al. (1984) performed a study on axial impact injury prediction to the head-neck system and found inconsistent correlation between peak impact force and observed levels of injury. The authors asserted that a parameter that would account not only for peak force, but also the duration of force, would likely be a better failure predictor. However, no such criterion exists for the prediction of serious neck injury. Other studies have been conducted to characterize axial loading response and stability of the spine (Pintar, 1989; Pintar 1990; Pintar, 1995b; Nightingale, 2000).

Gadd et al. (1971) also performed research on the response of the neck when subjected to singular loading conditions, which ultimately helped lead to the establishment of maximum tolerances for singular loading conditions. The study utilized cadavers subjected to extension and lateral flexion loading and determined a tolerance of 22.6 N-m bending moment before evidence of minor neck injury was found. Other researchers have used singular loading analyses to develop injury prediction tolerances and advance the understanding of neck kinematics in crash situations (Mertz, 1967; Severy, 1955; Macnab, 1964; Careme, 1989; Nusholtz, 2000; Deng, 1998; Matsushita, 1994; Thunnissen, 1995; Wismans, 1986; Wismans, 1987; Kallieris, 1990; Cheng, 1982; Panjabi, 1991; Panjabi, 1998a; Panjabi, 1998b; Grauer, 1997; Winkelstein, 1997; Oda, 1991; Oda, 1992; Penning, 1992a; Penning, 1992b; Ewing, 1972).

3.3. Combined Loading

It is considered unlikely that a subject will be subjected to such a "perfect" singular load described above. More likely, the subject will experience a loading condition that is a combination of these forces and moments, requiring an injury criterion that accounts for multiple loads. Two injury criteria have been developed to account for combined loading of the neck: N_{ij} and Modified N_{ij}.

3.3.1. N_{ij}

The National Highway Traffic Safety Administration expanded on the singular loading criteria with a new neck injury criterion, N_{ij}, as it was acknowledged that the singular loading criteria did not account for the potential combined effects of simultaneous neck axial loading and bending moments (Kleinberger, 1998). This criterion is based upon a linear combination of loads and moments. Proposed limits for the N_{ij} criterion were originally scaled for all ATD sizes based on limits determined for the 3 year old child ATD (Kleinberger, 1998; Mertz, 1997). An update of the critical limits was performed the following year for the 5th percentile female and 50th percentile male ATDs in tension and compression to reflect data gathered from cadaver testing (Eppinger, 1999). Based on feedback from the Alliance of Automotive Manufacturers and the Association of International Automobile Manufacturers, limits were adjusted again in March, 2000 (Eppinger, 2000). Final rulings for N_{ij} limits set peak values for axial loading of the neck in conditions where little or no bending was present, and also included an additional load factor to account for muscle tension absorbing some of the impact load (Eppinger, 2000; Mertz, 2000).

The N_{ij} criterion addressed the problem of compound loading conditions of the neck. The equation used to calculate N_{ij} is given below in Equation 1.

$$N_{ij} = \frac{tension\ (F_z)}{critical\ tension\ (F_{zc})} + \frac{flexion\ moment\ (M_y)}{critical\ \ flexion\ moment\ (M_{yc})} \qquad \textbf{Equation 1}$$

In this equation, F_z is the axial load of the neck, in either tension or compression. The M_y value is the moment within the sagittal plane, in either flexion or extension. The denominator values, F_{zc} and M_{yc} are the critical values for axial load and moment of the neck (Table 7). It is calculated using the simultaneous time histories of the neck tension and moment.

Table 7 – Critical N_{ij} Intercepts as Measured in the ATD (Eppinger, 2000)

Symbol	**Loading**	**95% Male ATD***	**50% Male ATD**	**5% Female ATD (In-Position)**	**5% Female ATD (Out-of-Position)**
F_{zc}	Compression (N)	8216	6806	4287	3880
F_{zc}	Tension (N)	7440	6160	3880	3880
M_{yc}	Flexion (N-m)	415	310	155	155
M_{yc}	Extension (N-m)	179	135	67	61
F_{max}	Peak Tension (N)	5030	4170	2620	2070
F_{max}	Peak Compression (N)	4830	4000	2520	2520

***** 95% Male ATD is not included in final ruling, but performance limits were given for informational purposes

Once N_{ij} is calculated, it is compared to a maximum allowable value of 1.0. If this value is exceeded, then the neck loading is too great and the potential for injury exists. It should be noted that the final ruling for critical values in N_{ij} also includes separate individual peak tension and peak compression values. This is because it was determined that the maximum amount of allowable tension or compression was too great in situations where there were zero or low moment values. For this reason, maximum values are given to limit the maximum tension and compression values to more stringent requirements (Eppinger, 2000) (Table 7) (Figure 1).

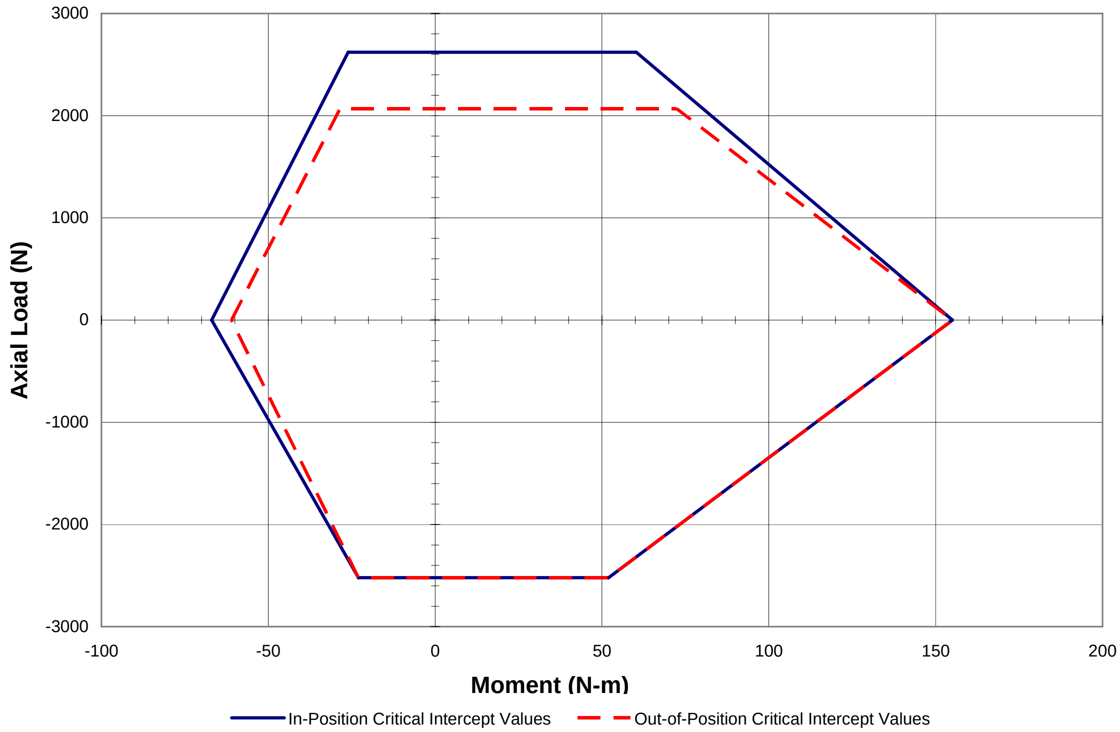

Figure 1 - N_{ij} Criteria for 5th Percentile Female ATD (Eppinger, 2000)

Differences in initial position are also distinguished in the N_{ij} criteria. Out-of-position values are given for the 5th percentile female ATD and have lower critical levels than in-position criteria (Table 7) (Figure 1). This is because it was speculated that an in-position occupant would be aware of an impending collision and would stiffen muscles, enabling the muscles to carry some of the load from the collision (Eppinger, 2000). Currently, there are no out-of-position values in the final N_{ij} ruling for any ATDs except for the 5th percentile female ATD.

The N_{ij} criterion was developed with the intent to assess serious neck injuries (AIS3+) in frontal impacts with deploying airbags (Schmitt, 2001). This is similar to the type of impact scenario in which a parachutist or aviator wearing HSM would be most at risk, where the head and neck are forced into flexion-extension at the onset of the acceleration. Therefore, it is expected that N_{ij} will be an appropriate starting point for the development of an injury criterion related to the neck forces and bending moments induced by HSM.

3.3.2. Modified N_{ij}

Duma et al. (1999) proposed that the N_{ij} injury criteria could be augmented by allowing for the resolution of bending moments both in the sagittal plane and the coronal plane. Based on studies of head and neck interaction with side-mounted airbags, they proposed a new, modified N_{ij} equation as shown in Equation 2.

$$Modified\ N_{ij} = \frac{tension\ (F_z)}{critical\ tension\ (F_{zc})} + \frac{bending\ moment\ (\sqrt{M_y{}^2 + M_x{}^2})}{critical\ \ flexion\ moment\ (M_{yc})} \quad \textbf{Equation 2}$$

In this equation, F_z is the axial load of the neck, in either tension or compression. The M_y value is the flexion moment within the sagittal plane. The M_y value is the lateral flexion moment. The denominator values, F_{zc} and M_{yc} are the critical values for axial load and flexion moment of the neck (Table 7). This is not intended to be used in cases with extension.

The intent of this equation is to provide a conservative basis for neck injury risk evaluation for lateral bending of the neck, such as in use for side impact injury evaluations. This is based on several assumptions; that the spinous process and vertebral arch are not load bearing in either flexion or lateral bending conditions and that the vertebral body, ligaments and muscles are similar in their load bearing characteristics both for frontal and lateral bending (Duma, 2000; Duma, 2002). Duma et al. (1999) conducted experimental studies with the HIII infant, where injury risk was calculated with 3 different methods. N_{ij}, N_{ij} oriented in a lateral direction and Modified N_{ij}. It was concluded that the Modified N_{ij} value offered the best correlation to the impact simulation, since it accounted for bending of the neck out of the sagittal plane as well as in the sagittal plane.

Just as N_{ij} was developed to account for compound loading conditions of the neck, and the corresponding lower failure threshold when subjected to compound loading that researchers have seen (Kleinberger, 1998), Modified N_{ij} accounts for multiple bending directions of the spine. It is believed that bending out of the sagittal plane would contribute to failure as much as bending in the sagittal plane. Further work to validate this criterion is needed, but research has been performed to aid our understanding of bending tolerance in the lateral direction. Gadd et al. (1971) performed a series of experiments on cadavers to determine load and flexion response of the neck in multiple directions. The conclusion of their study was that lateral bending stiffness was stiffer than hyperextension bending stiffness, though results in both directions were similar. Other researchers have performed studies with special attention to head-neck dynamics in lateral bending; however, no one has proposed a new criterion to account for this lateral bending (Wismans, 1984, Wismans, 1986; Schneider, 1975; Ewing, 1978a).

As the Modified N_{ij} criterion was developed to account for out-of-plane bending of the head-neck, and is essentially identical to N_{ij} for in-plane bending, it is felt that Modified N_{ij} would likewise be a useful starting point for development of an injury risk criterion for HSM induced loading. Accounting for out-of-plane bending will be especially important if studies are conducted on lateral impacts or other scenarios where large lateral bending moments are expected.

3.4. Minor Extension/Flexion

Due to the prevalence of minor neck injuries in auto accidents, particularly rear-end collisions, considerable effort has been spent to identify injury mechanisms and develop injury criteria for less severe, AIS1 type injuries. These injuries are hypothesized to be predominantly soft tissue injuries, where no specific radiographic evidence of injury exists after an event that may result in symptoms of aches and pains to the crash victim. Due to the multitude of hypothesized injury mechanisms, separate criteria have been developed, each with the intent of being able to predict injury based on its associated injury mechanism. Of the three types of criteria discussed below, NIC is said to be valid only up to the maximum extension phase of the neck. IV-NIC is defined for evaluation under both flexion and extension phases separately. N_{km} was developed to provide injury risk evaluation for the entire crash event, calculating risk of injury during the event irrespective of flexion or extension motion of the neck.

An excellent summary of the biomechanical mechanisms that are hypothesized to cause these minor injuries, and that led to the development of the discussed injury criteria, is provided by Yoganandan et al. (2002b). Other studies also discuss the physiology and biomechanics of "whiplash" injuries (Hell, 2002; Barnsley, 1994; Sturzenegger, 1994; Evans, 1992; Bogduk, 1984; Bogduk, 1986; Schrader, 1996; Galasko, 1996).

An important anatomical detail has recently been the focus of several studies examining rear end neck injury mechanisms. The role of the cervical facet joint during rear impact dynamics was studied by Yang et al. (1997). They found that as axial compression of the spine increased, the shear stiffness of the spine decreased. It is speculated that this is because as the spine is compressed, cervical ligaments can loosen. This would make it easier then for shearing displacement of the vertebrae and thus shear-induced soft tissue injuries of the neck. They hypothesize that this is why there is a marked difference in the occurrence of minor neck injuries in rear end impacts (where the neck is first in compression and extends rearward) than in frontal impacts (where the neck is first in tension and translating forward). Other studies have also been performed to investigate the role of the cervical facet joint in neck injury mechanisms and to better define kinematic motion during rear-end impacts (Stemper, 2001; Doherty, 1997; Emori, 1990; McConnell, 1993, Winkelstein, 2000; Kumaresan, 1998; Panjabi, 1993; Ono, 1999; Ono, 1997a; Ono, 1997b; Ono, 1993).

3.4.1. NIC

A new criterion was developed by Bostrom et al. (1996) based on an injury mechanism proposed by Aldman (1986). Based on Aldman's hypothesis, Bostrom et al. (1996) developed the Neck Injury Criterion (NIC) to account for a pressure effect in the neck caused by cerebro-spinal fluid flow during the rear-end impact event, and to predict AIS level 1 neck injuries for crashes at 20 km/hr or less (Bostrom, 1996; Bostrom, 1997). The formula for the criterion is shown as Equation 3.

$$NIC = a_{rel} \times 0.2 + v_{rel}^{\ 2} \qquad \textbf{Equation 3}$$

In the NIC, a_{rel} is defined to be the acceleration difference between T1 and C1, while v_{rel} is the velocity difference between T1 and C1. The 0.2 value is length parameter expressed in meters, an approximation of the length of the neck. The recommended injury tolerance limit is a maximum of 15 m^2/s^2 (Bostrom, 1997).

The principle on which this criterion is founded is based on the head motion during the initial phase of rear-end whiplash loading (i.e. the extension phase). In this initial phase, the head translates backward, putting the upper spine in local flexion. As the head rotation catches up with neck rotation, the head and neck begin to extend backwards, resulting in a transition from flexion of the spine to extension. The entire time for this full motion is said to be 150 ms. Therefore, by definition, NIC is defined to be the 3 ms maximum calculated from the first 150 ms of the crash event. It is speculated at this point of maximum extension, a large pressure gradient is developed in the cerebral spinal fluid, as the velocity and direction of the fluid flow may change as a function of the rapid volume change of the spinal cavity (Bostrom, 1996; Svensson, 1993). This pressure gradient may be enough to cause damage to surrounding nerve ganglion, causing a low-severity, AIS1 type, injury (Bostrom, 1996; Bostrom, 1997).

Darok et al. (2000) performed testing on several PMHS in an effort to validate the NIC. Testing on PMHS used pressure transducers to correlate increased cerebral spinal fluid pressure to increasing values of NIC. It was found that NIC was correlated with the magnitude of the pressure of the cerebral spinal fluid in the subject. The authors also conducted testing on volunteers in low severity simulations. In tests where the volunteers were exposed to impacts corresponding to NIC values of 8 or less, no complaints of minor injury were received. At NIC values of 10, minor complaints of muscle pain, limited neck motion and headaches were received. The researchers found NIC to be very well correlated to the impact parameters of velocity change and crash pulse and it displayed good agreement to the subject's head angulation, head angular acceleration, neck torque and head and torso accelerations. In practice, the authors found measurement of the original parameters of C1 and T1 prohibitively difficult to measure. For test efficiency and reliability the original NIC formula was modified in that C1 acceleration was replaced by head acceleration and T1 acceleration was replaced by torso acceleration. Also, instead of using NIC at exactly 50 mm of head retraction, the NIC was said to be that calculated the 3 ms maximum NIC from the first 150 ms of the event. Other research has been performed to generate data that could be used to validate this rear-end injury criteria, but did not diverge significantly from what has previously been discussed (Bohman, 2000; Zuby, 1999; Wheeler, 1998).

One of the main drawbacks of NIC is that it is only valid within the first 150 ms of a crash (Schmitt, 2001; Bostrom, 2000). This is due to the fact that the hypothesis on which the injury criterion is based, predicts the injury occurs just before or at the time of maximum extension of the head-neck system, which occurs in approximately the first 150 ms of a low severity crash. This does not account for additional loading of the neck as the crash event continues. It cannot be used to predict injuries caused by hyperextension or other injury mechanisms (Darok, 2000). Therefore it is recommended to use this criterion in tandem with other criteria, such as maximum neck loads to accurately predict the risk of injury. Finally, as mentioned previously, this criterion is only intended, and has only been validated for, low velocity rear-end collisions where the AIS level is 1 (Bostrom, 1997). For this reason, NIC should only be applied to HSM studies involving low severity loading.

3.4.2. IV-NIC

A new neck injury criterion, Intervertebral Neck Injury Criterion (IV-NIC), was proposed by Panjabi et al. (1999) that injury occurs when intervertebral rotation exceeds its physiological limit during whiplash. This criterion was developed as the authors concluded that "no definitive correlation was made between NIC values and actual clinical symptoms in whiplash trauma." The intent of this criterion was to allow for the evaluation of the cervical vertebrae motion at every intervertebral joint, thus creating a local neck injury criterion instead of a more global injury criterion based on a few kinematic measurements. The equation for the intervertebral neck injury criterion is shown in Equation 4.

$$IV\text{-}NIC_i = \frac{\theta_{trauma,i}}{\theta_{physiological,i}} \qquad \textbf{Equation 4}$$

In this equation, $\Theta_{trauma,i}$ is defined as the amount of intervertebral motion that occurs during the whiplash event. The $\Theta_{physiological,i}$ is defined as the maximum physiological limit of intervertebral motion. The risk of neck injury is defined when the ratio of IV-NIC is greater than one, indicating that the intervertebral motion exceeded normal physiological limits during the whiplash event (Panjabi, 1999). Practical use of IV-NIC would be extremely difficult. IV-NIC would have to be calculated for every intervertebral joint in the cervical spine, making the amount of data acquisition required for proper implementation very elaborate. Additionally, it would be required that the maximum physiological range of motion for each intervertebral joint be known prior to whiplash testing. Ivancic et al. (2004) proposed injury thresholds for maximum IV-NIC at each intervertebral level for frontal impact with muscle force replication. During this detailed analysis of IV-NIC Panjabi et al. (2004) measured the highest peak strains at C3-C4 forcing that region to be at the greatest risk for injury during a frontal collision. Due to the complication that this criterion is based on the total range of allowable motion on each test subject (in effect a fully customized injury criterion) it is thought that IV-NIC is not currently suitable for neck injury predictions.

3.4.3. N_{km}

Schmitt et al. (2001, 2002) assess that N_{ij} is not appropriate for use in low-speed rear-end collisions as it was developed specifically for serious (i.e. airbag deploying, AIS3+) frontal collisions. Similarly, the authors felt that a more informative criterion could be developed than NIC, since NIC is only useful for rearward neck retraction in the first 150 ms of a rear impact. In order to better evaluate neck injuries under an entire low-speed rear-impact scenario, they developed a new injury predictor, N_{km}, which is based on shear and flexion-extension bending moments. This seems to correlate well with the low speed injury mechanism proposed by Yoganandan et al. (2002a).

Yoganandan et al. (2002a) hypothesized two injury mechanisms for a whiplash type injury. The first was during the initial phase of impact, as the upper cervical spine was in flexion (due to the head lagging the neck response). The hypothesis is that this caused a stretching of posterior neck muscles and ligaments, which may be linked to sub-occipital pain. Then, as the head response caught up with the neck, it went into full extension. During extension, the lower spine facet joint (C5 to C6) experiences different loading conditions on the anterior and posterior sides, which can also lead to sliding of the facet joint surfaces relative to one another. This can result in a pinching mechanism and neck pain. According to Yoganandan et al. (2002a), electromyograph signals show that in volunteer rear-impact testing, initial muscle activation does not occur until 100 ms and that full muscle contraction does not occur until 150 to 170 ms into the event. At this point, the authors claim, the neck has already fully extended.

Since both of these injury mechanisms occur before or at maximum head-neck extension, Yoganandan et al. (2002a) believe that muscle contraction is not primarily associated with head-neck injury risk in a crash situation. However, as they indicate, this would seem to be the case where the occupant is unaware of impending impact. Where impending impact was noted by the occupant, they may have ample time to tense neck muscles before the impact occurs, possibly changing and limiting neck motion under the impact event.

Based on the injury mechanism hypothesis of Yoganandan et al. (2002a) and the experimentation of Schmitt et al. (2001), a new neck injury criterion was introduced to assess the forward kinematic phase of a rear-end collision. The N_{km} injury prediction criterion was proposed as shown in Equation 5 (Schmitt, 2001; Schmitt, 2002).

$$N_{km}(t) = \frac{F_x(t)}{F_{\text{int}}} + \frac{M_y(t)}{M_{\text{int}}} \qquad \textbf{Equation 5}$$

In this equation, F_x is the shear force on the neck. The M_y value is the flexion or extension bending moment. The N_{km} criterion is basically a modification of the N_{ij} frontal impact criterion, with specific intent to be used in low-speed rear-impact scenarios. New intercept values were determined in place of the normalizing critical force and moment values used for N_{ij} (Table 8).

Table 8 - Intercept Values for N_{km} Using TRID or BioRID ATD (adapted from Schmitt, 2001)

Load Case	Value
Extension (N-m)	47.5
Flexion (N-m)	88.1
Positive and Negative Shear (N)	845.0

The new criteria was validated against a series of 40 sled tests using the TNO Rear Impact Dummy (TRID) and Biofidelic Rear Impact Dummy (BioRID) ATDs, and it was concluded that N_{km} allows the inclusion of forward motion (i.e. flexion phase) of the neck for low severity crash injury prediction (Schmitt, 2002). Due to differences in the mechanical response of the TRID and BioRID ATDs, the authors note that different intercept values may have to be determined for each ATD. Presently, there are no experimental results to evaluate the Hybrid III ATD with the N_{km} criterion. Correlation with N_{km} predictions and actual minor neck injury has not yet been provided in the literature.

3.5. Injury Criterion Summary

In an effort to determine the usefulness of the various neck injury criteria on experimental tests, Yoganandan et al. (2000) performed testing on 5 PMHS: 4 small females and 1 large male. Rear end impacts were simulated using a sled from between 4.3 m/s and 6.8 m/s. For all tests, the NIC limit of 15 m^2/s^2 was exceeded, indicating a high risk of AIS1 injuries. Other criteria (N_{ij}, peak extension, and peak tension) predicted a likelihood of more serious AIS3+ injury. No conclusions were drawn as to the suitability of one criterion versus another; however, N_{ij} predicted a 22% injury risk for one specimen, where the specimen received a fracture of the 5th vertebral body, and hematomas in the C1-2 and C5-6 facet joints. It is important to note that N_{ij} does not predict the location of expected injury, only a global assessment as to the probability of an injury being sustained. N_{ij} also predicted a zero and less than one percent chance of AIS3+ injury in all other cases, where the specimens ranged from no injury to disk or ligament rupture. The study performed by Yoganandan indicates that the testing was severe enough to cause minor whiplash trauma to all specimens, and in the one case where serious injury was experienced, N_{ij} predicted an appropriately higher risk for injury than the other testing scenarios. For its applicability regarding HSM studies, use of the NIC is questionable for high severity impacts, since its limit was always exceeded in low severity impacts involving no HSM. However, it still may be useful to calculate NIC values for low severity HSM impact cases, to gauge the additional risk that HSM poses on the NIC scale.

For accurate assessment of risk from varying mechanisms, it has been demonstrated that calculation of injury risk must be performed using several different criteria. Multiple neck injury criteria have been developed to predict neck injuries in both minor (AIS1) and more serious (AIS3+) impacts.

For assessment of serious injury mechanisms (AIS3+), single loading criteria do not sufficiently account for complex loading scenarios of the neck; rather, N_{ij} would be utilized to account for both axial loading and bending moments of the neck. The Modified N_{ij} value accounts for effects of lateral bending of the neck which otherwise would not be considered in the traditional calculation of N_{ij}. If bending occurs only in the sagittal plane, the Modified N_{ij} value will give the same prediction of injury risk as N_{ij} itself. Therefore, it is recommended that Modified N_{ij} may be used to assess injury potential for AIS3+ level injuries in flexion or lateral flexion cases, and that singular loading values be monitored to ensure they do not exceed their individual limits.

The Neck Injury Criterion, or NIC, was specifically developed to predict minor neck injuries (AIS1) for low-severity impacts. This will not be very useful for high-severity HSM studies where severe injuries are likely, but it should be considered for low severity impact cases. IV-NIC is not fully defined for actual use as a valid neck injury predictor, and it would be extremely difficult to implement as it is based on each test subjects' total range of intervertebral motion. At present, no method to accurately access this value is available. As it is an adaptation of the N_{ij} formula with the intent to be used in low-severity situations (AIS1), N_{km}, like NIC, is not expected to be useful for predicting injuries at high-severity impacts; however, it can be calculated along with NIC for comparison during low-severity studies.

A final note regarding injury criteria is that the use of appropriate limits for the specific test subject is extremely important. Testing done by researchers that led up to singular measurand limits was performed on human cadavers and pediatric pigs. The same limits for human or animal tissues may not necessarily be applicable when the input force is applied to ATDs. Much effort has gone into determining and validating performance requirements for ATD head-neck systems and in determining accurate load thresholds for ATDs that correspond to mechanical limitations for humans (Wismans, 1983; Culver, 1972; DeSantis, 1991; Prasad, 1997; Seeman, 1986; Walker, 1973; Linder, 2002; Eriksson, 2002; Viano, 2002; Cappon, 2001; Siegmund, 2001; Kim, 2001). Stemper et al. (2004) established corridors, not injury thresholds, of the human head-neck response for rear impacts with the intent of using the data to test validity of an ATD response. It is extremely important that appropriate limits for injury criteria be used when calculating injury risk. Criteria values are most often specified for human subjects and for specific ATDs and the criteria are not interchangeable. Outputs from the use of Van der Horst's model, for instance, will be in the form of data representative of a live human occupant, not a simulated ATD. It will be important, to utilize realistic criteria for N_{ij} critical values, as the currently accepted criteria is for use on ATDs.. Unfortunately, these corresponding human critical limits are not readily available.

4. Head-supported Mass Studies

4.1. Head-supported Mass

The phenomenon of HSM-related injuries has been an issue since the advent of hard-shell helmets in the 1950's, and was exacerbated by the introduction of night vision goggle systems and other helmet borne equipment in the 1980's; however, data are still very limited when it comes to characterizing the specific risk of injury when subjected to impact scenarios (McEntire, 1997). Research in the area of HSM can be divided into four main categories: retrospective studies where researchers have polled existing data to draw conclusions regarding the effects of HSM, experimental work characterizing the physical performance of those encumbered with HSM configurations, experimental work using volunteers and ATDs to determine weight limitations for HSM to minimize injury risk, and computational simulations where computational models were used to predict the behavior of occupants based on given input parameters of acceleration environment and occupant parameters, including HSM configuration.

4.2. Retrospective Studies

Many attempts have been made by researchers to quantify the risks of cervical spine injuries for aviators. It has been hypothesized that there is an increased risk for neck injuries as a function of the weight of the aviator's helmet, the subsequent shift of the center of gravity of the head-helmet system from the head center of gravity, and multiplication of that weight and center of gravity shift by the g-forces, or dynamic shock, to which the aviator is subjected (Butler, 1992). Methodology for measuring helmet-system mass and center of gravity location is detailed by Shipley et al. (1993).

In a review of new technologies available for military aviator helmets, Rood and du Ross (1998) describe the problem of adding functionality to the aircrew helmet and how this pushes biomechanical safety of the pilot in the wrong direction. However, they offer only a study of impact, fit and noise attenuation properties of upcoming helmet technologies versus in-service helmets. No attempt is made in their report to quantify biomechanical safety limits for HSM, except that a reduction in mass and moments of inertia would be desirable with similar or better impact crash protection from new helmet systems.

Shannon and Mason (1997) concluded a 10 year retrospective database study to determine the injury rates of U.S. Army aviators involved in accidents and the relationship to wearing NVGs. In their study, data were analyzed from 357 rotary-wing mishaps affecting 704 crewmembers from 1985 to 1994. The data were analyzed in several different ways and provided insight into the likelihood of injury with aviators' use of NVGs. As data were pulled from the U.S. Army's Automated Safety Management System (ASMIS), they contained information regarding the U.S. Army's classification of the accident as survivable, partiallysurvivable or nonsurvivable. It was found that nonsurvivable mishaps incurred an 87% likelihood of head-neck injury while survivable mishaps saw only a 19% chance of head-neck injury. However, after data are controlled for the survivability of the mishap, it was found that NVG users experienced a 45% greater chance of head-neck injury compared to non-NVG users (Shannon, 1997).

It was also found that the two different types of NVGs in service during the time of the study produced different risks of injury. Aviator's Night Vision Imaging System (ANVIS) goggle users were characterized as having a "higher, but nonsignificant, risk of head or neck injury compared with non-NVG users". On the other hand, the crewmembers wearing AN/PVS-5 goggles experienced a 162% greater likelihood than non-users to experience head or neck injury. This difference was assumed to be correlated with the ANVIS goggles' breakaway feature on the helmet mount (Shannon, 1997). While these data suggest a higher risk of injury with NVG use in general, it also was found that horizontal impact velocity for NVG injury incidents was higher than horizontal impact velocity for non-NVG injury incidents. The authors suggested that mission profiles for NVG users may be different than for non-NVG users (i.e. more "nap-of-the-earth" flying). Given the nature of this type of flying, these flight profiles may be more prone to accidents in general, therefore exacerbating the problem of NVGs and neck-related injury.

Injuries to the upper cervical spine are also known to be attributed to pure acceleration loading. Jones et al. (2000) studied upper cervical neck injuries experienced among US fliers of high-g capable attack aircraft. It was found that there were two points in the careers of aviators at which they were susceptible to sustaining neck injuries. First, early in their careers when the aviators had yet to fully learn techniques to avoid neck pain. The second most likely time to sustain injury was after the pilot had logged >1,000 flight hours. This was attributed to the cumulative "effect of many load-bearing events takes its toll on the cervical disks, as the pilot ages, the supporting ligaments, muscles, and disc structure itself are less resistant to injury" (Jones, 2000). This study opens the question of what those physiological changes are as the pilot accumulates more flying time.

Hamalainen et al. (1993, 1994) attempted a study to identify factors which may make aviators likely to experience neck pain. In a study of student fighter pilots, the authors took body measurements of pilots, including neck circumference and head weight, and measured isometric neck strength. On the follow up study the authors noted that no dissimilarities were apparent between groups that had experienced neck pain and those that did not experience neck pain. It was concluded that higher neck strength could not yet be correlated with a decreased risk of injury, and also that no body measurement would help to pre-screen fliers for susceptibility to neck pain or neck injury.

However, radiological evidence has been used to both screen candidates to identify those most at risk to neck injury, as well as to diagnose injuries after their occurrence. Spinal X-ray screening was used by the Royal Norwegian Air Force to screen fighter pilot candidates for preexisting susceptibility for spinal injury (Anderson, 1991). This new screening process was added to the medical evaluation due to the growing prevalence of neck injury, in particular cervical spine injury, among fighter pilots due to the high loads that the helmet and other equipment add during maneuvering. It is noted that the value of interpreting existing pathological evidence as a higher risk for injury is not certain.

A case study of a neck injury sustained during a parachute jump stresses the importance of proper initial positioning and low helmet mass in avoiding injuries (Makela, 1997). A severe neck flexion moment was experienced during a parachute deployment and the soldier subsequently experienced severe neck pain. Radiological evidence showed that degeneration and protrusion was seen in the C5 through C7 intervertebral disks.

Surveys that are intended to identify techniques employed by pilots to reduce the incidence of neck pain or neck injuries have revealed ways to lesson the risk of sustaining injury. F/A-18 pilots of the Royal Australian Air Force were questioned regarding techniques that they used for head positioning to minimize neck pain or injury (Newman, 1997). It was found that if the head were braced prior to application of high g-loading and not moved during the maneuver, the risk of sustaining an injury was lessoned; however, the specific benefit that bracing techniques offer have not been quantified. Jones et al. (2000) speculated that neck strengthening exercises could help to improve the odds against receiving a cervical neck injury. Unfortunately, the inconsistencies of the exercise methods for neck strengthening used by the respondents in his study meant that no conclusions could be drawn as to whether this mitigated risk of injury or not. In a related study on F-16 pilots, it was found that neck strengthening exercises and bracing the head prior to maneuvers reduced risk of injury (Albano, 1998). It was also noted that the risk of sustaining a neck injury increased by 6.9% for every 100 hours of flight time, suggesting increased risk of injury due to degeneration or fatigue due to long term exposure to high g-loading.

4.3. Experimental Studies

Several experiments have been performed with human volunteers and ATDs to characterize the biomechanical behavior of the neck while wearing HSM. Experiments have been performed where accelerations are measured and governing behavior of the subjects are characterized, in an attempt to determine appropriate limitations for HSM relative to aviator safety.

4.3.1. HSM Performance Studies

Glaister (1987) performed a study on head mobility with HSM when subjected to high g-loading. This testing was not designed to quantify injury risk, but to evaluate the ability of aviators to remain able to control their head movements. It was found that an unloaded head was mobile at 6 Gz; however, the ability to maintain head mobility dropped to 4.5 Gz to 5.5 Gz when subjected to loadings of 2 lb to 4 lb of head-supported mass. No impact testing was performed to evaluate potential limits for injurious levels of HSM.

Watkins et al. (1991) created a mathematical model that utilized experimental data from kinematic tests of volunteers with HSM in order to predict head-neck kinematics of the subjects. The validated model can be used to extrapolate kinematics data for HSM configurations where volunteer testing is not safe, due to risk of injury. This model can be used to validate experimental and other computational models of head-neck response to various HSM loading conditions; however, the model does not offer a mechanism to predict the level of injury risk due to these configurations.

A series of tests was performed at the U.S. Air Force Research Laboratory to identify the modes of head-neck response to vertical impact and to determine the parameters of particular importance in developing a model that could accurately predict the response of the head-neck system to various HSM loads and impact conditions (Ziejewski, 1998). 125 experiments were conducted using 27 volunteers up to an impact of +10 Gz. Helmet mass and center of gravity was varied to simulate a range of available military helmets. It was found that determination of the linear acceleration of the head center of gravity and calculation of the head pitch allowed a unique characterization of the head-neck response to the loading environment. The purpose of this study was to determine important parameters to predict head neck response under impact loading. No efforts to develop criteria for safety limits of HSM were reported.

Butler (1992) performed a series of experiments on volunteers to characterize head-neck response while wearing HSM under whole body vibration. Butler studied male aviators with helmets of various masses and centers of mass. Aviators were subjected to whole-body vibration of axial swept-sine frequencies from 2 Hz to 17 Hz. He concluded that head pitch motion was the predominant motion under axial swept-sine whole-body vibration. Changes in the mass and center of gravity increased the head pitch response but did not change the resonant frequency of the subject. Butler felt that vibration of the head and helmet would result in inertial moments that would act to increasing the load on the neck and neck muscles as the body attempted to control these motions. He found that under vibration, posterior muscles of the neck act in head extension while anterior muscles work in conjunction with gravity to support the head in flexion. Posterior muscles were seen to work much more consistently to control head-neck motion than anterior muscles.

These experimental results led Butler (1992) to conclude that there were three main criteria that needed to be considered and understood in order to characterize the biomechanical effects of wearing a helmet: acceleration environment, duration of exposure, and the posture of the helmet user. Based on his analysis of these data, Butler derived an upper boundary for HSM of a moment of .83 ± .23 N-m relative to the atlanto-occipital complex. This was based on both myoelectric responses of neck muscles and head pitch measurements. As the head-neck moment relative to the atlanto-occipital complex increased beyond .83 N-m, the amount of measured head pitch acceleration significantly increased. This coincided with an increase in myoelectric response in posterior muscles to the limit of 5% maximum voluntary muscle contraction as an upper limit for long duration, static muscle loading. Butler also noted that there was no significant difference in head pitch acceleration response for the unloaded (no helmet) case compared to the helmets with a weight moment less than .83 N-m.

A follow on study was conducted by the United States Army Aeromedical Research Laboratory based on the results of Butler's (1992) study. The intent of this study was to analyze pilot performance degradation when subjected to various configurations of HSM and random vibrations for four hours. The test subjects were Army helicopter pilots who were asked to aim a light beam attached to their helmets at random targets. Researchers found that the reaction time of the pilot significantly increased when the weight moment of inertia was increased beyond .78 N-m (Alem, 1995). Given the results of Butler and Alem's studies, the USAARL has adopted for male aviators a criteria of .80 N-m for the weight moment of the HSM relative to head center of gravity (Barazanji, 1998).

Although these limits might be physiologically well-founded for male aviators, they left a question as to whether they were well-suited as an umbrella limitation to cover male and female aviators alike. Barazanji et al. (1998) undertook an experiment to characterize the behavior of female head-neck acceleration while subjected to whole body vibrations. These tests were performed on female non-aviators (so as to eliminate variability in the amount of exposure test subjects had with helmet systems). The study's goal was to find the analogous limit to head-supported mass weight moments for female aviators. Barazanji et al. (1998) found, just as Butler did with male aviators, that head pitch accelerations for females seemed to be the most sensitive in responding to changes in HSM. The authors' study of HSM on female aviators came to the conclusion that .65 N-m was the upper limit, where head pitch acceleration began to increase with larger weight moments. Other research has come to similar conclusions that women may be more at risk for neck injury then men, based on lower overall neck strength (Kroonenberg, 1998; Morris, 1999).

Another important issue to note was that initial posture plays a very important role in the risk of sustaining neck injuries. Butler (1992) found during his study of whole body vibration that even minor changes in posture could cause significant changes in weight moments. He found that controlled posture situations produced significant variation in head pitch acceleration compared to uncontrolled posture scenarios. Therefore, it must be known for the desired computer simulation, whether or not the subject is trying to maintain posture or is in an uncontrolled situation (i.e. trying to maintain visualization of a target or "along for the ride" in an impact situation). Butler supports this theory and recommends against using controlled posture when assessing the effect of HSM in situations when posture of the individual is uncontrolled. Other researchers came to similar conclusions regarding the effects of posture on head-neck response (Ewing, 1975; Ewing, 1978b).

4.3.2. HSM Injury Studies

Recently, research has been performed to specifically characterize injury risk for HSM. Most experimental work has been performed using ATDs, rather than volunteer testing. There are several reasons to consider performing HSM research with biofidelic ATDs. Since serious neck injury criteria (i.e. singular loading and N_{ij}) are correlated with maximum ATD loads, they allow direct interpretation of injury criteria. Also, they allow experimentation over a wide range of impact scenarios, without risk of injuring a volunteer.

Bass (2002) performed some preliminary testing involving the use of a Hybrid III ATD subjected to HSM. Test scenarios were run with a variety of HSM configurations where both the weight and the distance from the head center of gravity were considered. These initial studies showed that most scenarios exceeded the N_{ij} criteria of 1.0, except it was noted that higher neck stiffness would lower the predicted N_{ij} value, bringing N_{ij} to below 1.0 in some cases. Also, Bass developed and provided further validation on the experimental results with computational modeling using MADYMO's Hybrid III ATD. Excellent agreement was found with HSM weights of 1.75 and 2.5 kg; however, the model predicted an N_{ij} of 1.35 with 4.5 kg as opposed to the experimental value of N_{ij} found to be 1.55. Bass concluded that further studies should be performed using cadaver test subjects and a HSM injury criterion needed to be developed that included anterior-posterior neck shear forces. It was also identified as desirable to validate experimental results with a computational model.

A study was conducted by McEntire et al. (2002, 2004) with the intent to characterize neck loads seen by paratroopers as the loads of the parachute opening are transmitted to the soldiers. Specific interest was given to neck loading caused by the use of two different helmet systems. The first helmet was the standard airborne helmet, the second was a helmet modified to include a HSM load. Testing by actual parachute drops out of a UH-60 Black Hawk helicopter displayed that the flexion moment of the neck caused by the HSM helmet was significantly greater than that of the standard helmet (20.83 N-m versus 11.81 N-m, respectively). None of the neck force or neck bending moments measured exceeded singular loading criteria; however, N_{ij} values had not yet been calculated for the tests. NIC or other minor neck injury criteria were not calculated for this experiment.

Ejection events (+Gz impact loadings) have been studied to identify the risk to aviator safety with HSM, including night vision goggles and other helmet mounted displays. Perry et al. (1993) conducted a literature review regarding helmet mounted system +Gz loading and, for an Advanced Concept Ejection Seat (ACES II) ejection seat, concluded that a helmet system having a weight of 2.27 kg or less, and a center of gravity from -0.51 to 2.79 cm and 1.02 to 3.56 cm on the anatomic x and y axes, respectively, would not induce a higher risk of neck injury than operational helmets without integral vision systems. Experimental tests using helmet systems that met these criteria were conducted using an Advanced Dynamic Anthropomorphic Manikin (ADAM) in an ACES II ejection environment and found to produce neck loads below the injury threshold (Perry, 1994).

As a follow-on study, Perry (1997) conducted experimental testing on both male and female volunteers to +Gz impacts with HSM. Similar loads were developed by both males and females at the occipital condyles, except that it was found that males tended to translate their heads rearward under +Gz impact, while females translated their heads forward during the same event. It was concluded that females may be at greater risk for neck injury under +Gz loading, since they experience the same loads that the male aviators experienced and females have a 25% greater injury risk in dynamic impact environments, based on automobile accident statistics.

A series of +Gz impact tests were conducted on volunteers and ATDs to determine the effect of various helmet weights from 1.5 kg to 3.0 kg on head-neck response (Buhrman, 1994). From the experimental results for the volunteer tests, neck compressive, shear and bending moments were determined by calculating head mass estimates with linear and angular accelerations. These results were then compared to the loads and moments measured in the ADAM ATD. It was found that the neck used in the ADAM ATD was in good agreement with loading calculated from the volunteer tests. These results then indicated that a 3.0 kg helmet at +10 Gz would not exceed injury limits developed by Mertz and Patrick (1971); however, at +15 Gz, injury limits were exceeded with a helmet mass exceeding 2.0 kg.

Shender et al. (2000) performed a series of tests to determine the effect of varying weight and center of gravity position of helmets, including several existing military helmet systems, on neck loading during simulated maneuvers. In this study, a solid aluminum neck was chosen over an ATD neck to eliminate unrealistic bending motion of the neck under loading. It was found that for helmet weights of 1.4 kg to 2.5 kg, a +12 Gz maneuver would introduce a neck load of 1010 N to 1112 N and a bending moment ranging from 78 N-m to 112 N-m. Since a solid aluminum neck was used in these experiments, it is unknown how the loads measured by Shender et. al would correlate to ATD maximum allowable neck loads.

4.4. Computer Simulations

While some of the experimental work also involved validation utilizing computer simulations, some research has been preformed solely with computer simulations. Researchers are using computer simulations to determine the convergence of increased HSM and aviators' risk of exceeding established injury thresholds for head-neck. The United States Army has utilized computational simulations to analyze kinematics in aviation accidents, often to assess the benefits of safety systems (Beale, 1996). Brozoski et al. (1998) performed a study using the Articulated Total Body (ATB) software package to simulate a variety of crash conditions for an AH-64 rotorcraft and the relationship to seat stroke and HSM configurations. The occupant simulated in the computer program was the mid-size male Hybrid-III manikin. Loads calculated on the manikin's occipital condyles were then compared to accepted injury thresholds for the Hybrid III to determine the likelihood of neck injury. Using this information, limits were developed to provide maximum mass and center of gravity locations for the crash sequences. Simulations used HSM of 0.45 kg, 1.35 kg and 2.70 kg, which were said to encompass the range of masses currently used by the US Army for HSM.

Conclusions from Brozoski et al.'s (1998) simulation were that with increased seat travel, the maximum amount of HSM to be safely worn by an aviator increases. If the seat is only able to stroke a minimum of 2.54 cm, then the maximum mass of headgear the aviator could safely wear in a vertical impact was determined to be 0.45 kg. If the seat is able to stroke the fully designed 25.4 cm, the maximum head-supported weight of 2.70 kg could be safely tolerated without increased risk of injury. However, in the horizontal impact scenario, the seat stroking distance did not play a role in the maximum HSM safely allowed. It was found that for a horizontal impact, the maximum amount of HSM that could be safely worn for all possible positions on the helmet is 0.45 kg. For horizontal impacts, no acceptable results were found for HSM of 2.70 kg and only limited safe positions were found for mass of 1.30 kg. In summary, the results of this study show that as HSM increases, the acceptable center of gravity location moves below and to the rear of the head center of gravity. Most importantly, the maximum allowable HSM was largely dependent on impact conditions, in particular the impact acceleration magnitude, pulse and shape (Brozoski, 1998). This will make it necessary to perform many different crash sequence scenarios to gain an appreciation for what the maximum safe level of HSM can be for an individual. The results of this testing suggest that a series of vertical impact tests and horizontal impact tests should be considered for the detailed head-neck HSM model to be validated.

Another computer model used to simulate the effect of HSM on military aviators was created using the ATB software package by Paskoff (2002). The intent of this model was to study the effect of HSM on cervical spine loads for 5th and 95th percentile male occupants during ejection scenarios. Conclusions from Paskoff's simulation were that the most dominant factor affecting head-neck kinematics was the resultant center of gravity of the head-helmet system, the total amount of head-supported mass and resultant moment of inertia were found to have a lesser effect on overall neck loads. Another computer model was created by Privitzer and Kaleps (1990), this model was also used to study the effects of HSM in aviation ejection scenarios, with similar results.

Race car drivers, like military aviators, are subjected to frequent high g-forces acting on their crash helmets. While racing, fatigue of the neck muscles begins to limit driver performance, and accidents under such high velocities can produce extreme motion of the head-neck system (Hubbard, 1994). The Head and Neck Support (HANS®) device was developed to help restrain the driver's head to reduce fatigue under normal racing conditions and to limit head motion in impact situations. A computer model was used to validate that the neck support could help to support the weight of the helmet under normal conditions and help prevent large motion in crash loading (Hubbard, 1990; Hubbard, 1994). While this may have little application in the military environment, it is gaining popularity in the civilian market.

4.5. Head-supported Mass Summary

Equipment added to soldiers' helmets can significantly affect head-neck dynamics under impact conditions. Conclusions have been made from retrospective case studies, experimental work utilizing volunteers and ATDs, and computer simulations that this new equipment poses an unquantified risk to soldiers. For this reason, further experimentation is necessary in order to identify realistic mass and inertial limits for new equipment designs, as well as to develop recommendations for preferred practices in order to minimize injury risk.

5. Muscle Activation on Neck Impact Response

5.1. Muscle Effect on Neck Response

Despite the fact that much experimental and computational work to characterize neck response in impact scenarios has been performed, the effect of musculature on head-neck whiplash-type response is still debated. This is not surprising, since the head-neck musculature is one of the most complex neuromechanical systems in the human body (Winters, 1990b).

Muscles can influence human head-neck behavior in two ways. First, the mere presence of muscle, even if it is not physically contracted, will add mass to the head-neck system and the muscle will have material properties that affect kinematics of the associated joints. Second, living beings have the capability to actively contract their muscle, either providing motion across a joint or joints, stiffening the bending properties of the joints, or both. Both passive and active muscle properties have been shown to influence the behavior of the head-neck system.

5.2. Passive Muscle

Muscle tissue accounts for the largest percentage of soft tissues in the neck (Siegmund, 2000). Therefore even when not actively contracted, the mass of the musculature and its mechanical properties will play a significant role in determining head-neck dynamics. Cadaver experimentation, of course, cannot be utilized to understand the role of active musculature, but experiments with PMHS's do offer a significant increase in biofidelity over ATD experimentation. Passive muscle effects can have a sizeable effect on the overall response of a PHMS.

Van Ee et al. (1998, 2000) performed experimentation with rabbits to identify changes in mechanical behavior of the test subjects with regard to freezer storage versus fresh tissue testing and the effects of mechanical conditioning of the muscles. The authors found that without pre-conditioning, muscle response was stiffer than live passive muscles, however, this response was not repeatable. Repeatability of results was made possible by pre-conditioning the muscles, but the pre-conditioning of the muscles also made them less stiff than live passive muscle. While freezing was found to have no effect on post rigor mortis muscle response, it was noted that a 61% decrease in failure stress was observed with post-mortem muscle as opposed to live passive muscle. These results suggest that cadaver experimentation also lacks some biofidelity beyond the known limitation of actual physical responses. To most accurately simulate live passive muscle response, results indicate that not preconditioning cadaver muscle may offer the most biofidelic response; however, pre-conditioning is necessary in order to obtain repeatable experimental results and for proper preparation of the osteoligamentous structures (Van Ee, 1998).

5.3. Active Muscle

The ability of the human reflex to activate neck muscles in time to have an effect on head-neck kinematics has been debated since the 1970's. Studies utilizing electromyography have been performed to determine the speed at which muscles can react under impact loading. Early testing led many researchers to conclude that cervical musculature would not react fast enough or generate force quick enough to provide support during severe impacts (Tennyson, 1977; Schneider, 1975). Due to this conclusion, it was considered acceptable that most computational models of the head-neck system do not include active musculature, as this was felt to have no bearing on the impact response. Recently, findings from experimental volunteer testing and computational modeling have begun to suggest that muscles do, in fact, react fast enough to affect the head-neck impact response.

To accurately simulate the central nervous systems' active muscle commands, response characteristics of musculature must be known. Butler (1992) recorded electromyograph activity of the anterior and posterior muscles of the neck for his testing of whole body vibration with head-supported mass loads. It was noted that the posterior muscles of the neck (splenius capitus and trapezius) showed bursts of activity synchronized with head and neck flexion. The anterior muscles of the neck showed little response to the sinusoidal whole body vibration in the sagittal plane. This led Butler to conclude that with HSM, the posterior muscles of the neck are most often used to stiffen the neck and influence the behavior of the head-neck system. This may also indicate that when encumbered with HSM, the neck muscles are in some form of muscle activation, and as a result, muscle response times may be even shorter than a relaxed neck that is unencumbered with HSM.

The time required to activate muscles is also crucial and needed to build a realistic time delay into the computer simulation. Magnusson et al. (1999) performed a series of tests on volunteers to determine the time delay from impact to neck muscle activation. They also took measurements to relate the onset of neck muscle activity to various events within the testing "event". The authors determined the average time delay from the onset of sled-acceleration to the beginning of neck muscle reaction was 112 ms, the time delay from initial trunk acceleration to the beginning of neck muscle activity was 94 ms, and the time delay from the onset of head acceleration to initial neck muscle activity was 59 ms.

Magnusson et al. (1999) noticed that neck muscles responded with differing time delays, and they speculated that with these time delays the neck muscles had different functions. The outer-most muscles, the levator scapulae, sternocleidomastoid, and trapezius muscles responded between 73 ms and 83 ms (measured from sled-acceleration onset). From the onset of head acceleration, these reaction times were from 13 ms to 22 ms. It was observed that these muscles, having the largest lever arm(s) and thus the largest effect on the gross motion of the head-neck, would act first to stabilize the spine. After the spine was stabilized, the inner layer of muscles, the semispinalis and the splenius muscles would react, at approximately 170 ms into the event. These primarily made small, finer adjustments to the positioning of the spine. It was concluded that since maximum head acceleration occurred approximately 50 ms after head acceleration began, the fast-reacting muscles would have an effect on the overall cervical spine response to an impact event. It was also noted by the authors that they found no difference in muscle response times for occupants expecting versus not expecting the impact. This was theorized to be related to the test methodology, where the trigger for neck muscle activity occurred as the backrest was accelerated into the subject's back resulting in a somato-sensory response for the onset of muscle tension.

Szabo and Welcher (1996) also conducted low-speed rear-end impact testing using volunteers, and measured neck muscle activity using electromyography. The test protocol was developed such that the subjects displayed relaxed posture muscle activity in the pre-impact state, indicating that they were unaware of the exact timing of the impact and were not braced prior to the sled pulse. They found that the onset of muscle activity occurred between 100 and 125 ms after initial bumper to bumper impact. This onset of muscle activity generally occurred before the neck reached full extension, and full muscle activation occurred at approximately 150 to 170 ms into the event, but after full extension. Unlike Magnusson et al., Szabo and Welcher found no clear differences in the activation times of the various muscles. However, this could be due to the fact that only the outermost layer of muscles were monitored for this study (anterior and posterior paracervical, trapezius and paralumbar), and Magnusson et al. found that the outermost layers of muscles reacted within the same 100 to 125 ms timeframe and the inner muscles reacted later (Magnusson, 1999; Szabo, 1996). Szabo and Welcher also noted similar reaction times to Magnusson with respect to the onset of head acceleration. They found that the reaction times of the outermost layer of cervical muscles were noted to be approximately 20 to 30 ms after the onset of head acceleration.

A study done by Reid et al. (1981) simulated whiplash loading by a sudden backward pull on the head. This direct acceleration of the head may lead to faster response times than in car-car impact simulation, if the stimulus for neck muscle contraction comes from head acceleration. This is backed up by the author's findings, showing that when subjects were instructed to resist as much as possible, some could react within 25 to 40 ms after the head tug. In other scenarios where the subjects were instructed to react as they desired, they displayed longer response times. This may be attributed to the fact that the tug was not severe (i.e. the subjects were relaxed and knew they wouldn't get hurt). Similarly, Brault et al. (2000) had determined that the muscle response time can be a function of the impact severity, thus with the low-severity impact, reaction times were slower. When instructed to react with maximum effort the subject reacted as if it were a severe loading situation to gain maximum muscle effort.

Experiments conducted by Ono et al. (1997) were intended to study vertebral kinematics in rear-end impacts. Electromyography was used to monitor the subject's neck muscle activity. It was found that volunteers could limit head kinematics by tensing their muscles, but it was also noted that volunteers tended to have relaxed muscle states as they became familiar with the testing. Other researchers have noted similar trends. Reid et al. (1981) noticed that once they became accustomed to the loading, the subjects could respond with much control. It is speculated that certain loading scenarios could be "trained" for to decrease the risk of injury in such circumstances. This would suggest that it would lessen the likelihood of injury to soldiers who would be subjected to repeated loading situations, if they were trained in a way that they would learn to anticipate the exact loading they would be subjected to.

In a more recent study by Ono et al. (2003) impact tests were conducted with both tensed and relaxed musculature of the volunteers. While the impact load was applied, the conditions of the neck for the two different muscle activation cases had no significant differences. However, there was an obvious difference in responses after the load was removed. The sternoclaidomastoid muscle (SCM) activated 50 ms after the impact for the tense condition but did not activate until 80 ms for the relaxed condition. This SCM activation suppressed head motions after the load was removed for the tensed subjects. Ono also compared the results of his tests to cadaver test data reported by Viano et al (2001). Noting that the impact load level was larger in the cadaver tests, the comparison describes the biomechanical responses caused by muscle conditions. A cadaver responds immediately in extension to the impact whereas a living human-being with relaxed muscles will respond briefly in flexion after 30 ms and then go into extension. Lastly, the cadaver can go beyond the biological limits of extension while the volunteer's head returns to the original position from extension. Ono (2004) reported that although the kinematics is different, the intervertebral motion was not significantly changed by the application of neck muscle tension.

Accident data have been investigated by researchers to infer the effect of neck muscle response on injury risk, based on the occupant's testimony, in particular their awareness of the impending collision. Olsson et al. (1990) found no difference in injuries to occupants of rear end collisions who were aware of impending impact and tensed their muscles prior to the collision and those who were not aware of the impending collision. It was noted, however, that none of the occupants who were aware of impending impact and pushed themselves against the seatback-head support was injured.

PMHS testing includes passive muscle effects only, thus leaving volunteer experiments as the only possible experimental setup with which to gain insight into muscle activation. Due to the fact that volunteers are used only in low severity simulations, the data collected are only validated for a limited range. Therefore to predict the effect of active muscles in severe impacts, computer models with simulated active muscles must be used. Van der Horst's detailed neck model uses accurately modeled neck musculature with capability to be actively controlled (Van der Horst, 2001; Van der Horst, 2002). It has been validated with low severity volunteer testing in frontal, rearward and lateral impact conditions. As was discussed in Chapter 2, during validation of Van der Horst's (2002) model, it was determined that low-severity type impacts required less active musculature for proper correlation with volunteer experiments. As the severity of the impact increased, the amount of necessary active muscle response increased to 100% for accurate simulation of volunteer responses (Wismans, 1998). As the severity increases beyond the safe limits of volunteer testing, it can be assumed that while the muscles may be overwhelmed by high severity impacts, they will also help to substantially stiffen the neck response, and will alter the possibility of neck injury.

Overall, it is presently accepted that neck muscle activation does occur sufficiently fast enough to influence the head-neck system under whiplash loading (Brault, 1998a; Brault, 1998b; Siegmund, 2000; Magnusson, 1999). Studies that report on the activity level of neck musculature under HSM loading, such as the one performed by Butler (1992), also create the question of whether the HSM encumbered neck muscles would actually respond faster, as they are already at a higher level of activation, than a relaxed vehicle occupant without HSM. It has been hypothesized that neck muscle activation under impact situations may actually contribute to the risk of other neck tissue injury or increase the risk of injury to the muscles themselves (Vasavada, 2002). This is due to the fact that as neck muscles contract, they induce further compression on the vertebral column, and as kinematic motion of the head continues, strain on the tensed muscle may increase.

5.4. Muscle Summary

In the 1970's, as researchers were attempting to characterize the head-neck response under impact loading conditions, it was felt (and thought to have been experimentally proven) that neck muscles did not react quickly enough, or with sufficient force to influence head-neck kinematics. Recent developments in the laboratory and in computational simulations suggest that muscle activation does develop sufficiently fast enough and with enough force to influence head-neck dynamics under impact situations. Therefore, for accurate computational simulation, it is important to choose a model that is capable of introducing active muscle behavior into a simulation.

6. Conclusion

Four factors were researched that encompass the issue of head-supported mass and its associated injury risk: computational models that can simulate the response of humans to a variety of impact conditions, neck injury criteria which are used to predict the likelihood of injury based on the response of a human or surrogate to impact loading, the effects of head-supported mass on head-neck kinematics, and finally the effect of muscle activation on human head-neck response. The results of this research as it relates to the head supported mass study did not change with the added literature from 2003 and 2004.

First, while many detailed computer models have been developed in the past decade for use in understanding neck injuries, none has been developed with as much detail and versatility as the MADYMO detailed head-neck model by Van der Horst (2002). This model has been widely validated in frontal, lateral and rear automotive type impacts and agrees reasonably with both PMHS and volunteer testing. Although not specifically designed for head-supported mass studies, it is the most validated biofidelic model available from which to begin a computational HSM study.

Second, neck injury criteria have been developed to predict neck injuries in both minor (AIS1) and more serious (AIS3+) impacts. In cases where single measurand loading does not sufficiently account for complex loading scenarios of the neck, N_{ij} could be utilized to account for both axial loading and bending moments of the neck. In addition, the Modified N_{ij} value accounts for effects of lateral bending of the neck which otherwise would not be considered in the traditional calculation of N_{ij}. If bending occurs only in the sagittal plane, the Modified N_{ij} value will give the same prediction of injury risk as N_{ij} itself. The Neck Injury Criterion, or NIC, was specifically developed to predict minor neck injuries (AIS1) for low-severity impacts. Although this will not be useful for high-intensity HSM studies, it should be considered for low-intensity impact cases, where low severity injuries are expected rather than high severity injuries. The Intervertebral Neck Injury Criterion, IV-NIC, is not fully defined for actual use as a valid neck injury predictor. IV-NIC would be extremely difficult to implement as it is based on each test subjects' total range of intervertebral motion. As it is an adaptation of the N_{ij} formula with the intent to be used in low-severity situations (AIS1), N_{km}, like NIC, is not expected to be useful for predicting injuries at high-severity impacts. However, it can be calculated along with NIC for comparison during low-severity studies.

Third, it has been determined that equipment added to soldier's helmets can significantly affect head-neck dynamics under impact and normal operating conditions. Conclusions have been made from retrospective case studies, volunteer and ATD experiments, and computer simulations that this equipment poses an un-quantified risk to soldiers. For this reason, further experimentation is necessary in order to identify realistic mass and inertial limits for new equipment designs, as well as develop recommendations for preferred practices to minimize injury risk.

Fourth, the role of neck muscle activation in head-neck kinematics has been a source of controversy for over 30 years. While in the 1970's data seemed to indicate that neck muscles did not react quickly enough, or with sufficient force to influence head-neck kinematics, recent laboratory findings support the idea that neck musculature can influence head-neck dynamic response. Future studies involving head-neck response should include provisions to investigate the role of neck muscles in preventing neck injury. Therefore, for truly accurate computational simulation, it is important to choose a model that is capable of introducing active muscle behavior into the simulations.

In summary, it has been shown that head-supported mass poses an unquantified risk to the safety of military personnel. Detailed computational simulations are available to accurately simulate human response to a variety of loading conditions, including simulations of live occupant response with passive or active muscles. Muscles have been determined to respond quickly enough in an impact to affect head and neck dynamics and lower the risk of serious neck injury. Neck injury criterion can be used to determine the susceptibility of the simulated occupant to potential neck injury. Further studies utilizing these tools will ensure that military personnel are not subjected to undue safety risks in the form of head-supported mass.

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