Virginia Tech - Wake Forest
Center for Injury Biomechanics

www.cib.vt.edu

CIB Report Number 2004-090

Analysis of Computational Neck Models

PREPARED FOR

Far Side Impact Research Committee
Attn: Ken Digges & Richard Morgan
FHWA/NHTSA
National Crash Analysis Center
G.W. Transportation Research Institute
44983 Knoll Square
Ashburn, VA 20147
Phone: 703-726-3543 Fax: 703-726-3530
Email: rmorgan@ncac.gwu.edu

PREPARED BY

Eric Kennedy, Joel Stitzel, and Stefan Duma
Center for Injury Biomechanics
114 Randolph Hall
Blacksburg, VA 24061
Phone: 540-231-3945 Fax: 540-231-2953
duma@vt.edu

June 4, 2004

Abstract

This report presents a literature review of the latest developments in the computational modeling area of neck biomechanics for specific application to the study of carotid artery injury in far-side impacts. First, mature computer models of the head and neck are reviewed to determine the most biofidelic and versatile model available. Second, full body computational models of the human are evaluated with specific emphasis on the neck region. Due to the requirements of detailed geometry for the cervical spine and neck, as well as validation in lateral impact scenarios, two models are recommended for potential use in studying far-side impacts and carotid artery injuries. The Total Human Model for Safety (THUMS) by Iwamoto et al. (2002) and the MADYMO Detailed Neck Model by Van der Horst (2002) both offer reasonable biofidelity and the accurate geometry required for studies into the lateral impact response of the neck. However, neither model offers full modeling of the soft tissues of the neck, requiring additional materials to be added to the existing models or extraction of pertinent information from these models to be used as input parameters for a more detailed local model of neck soft tissues.

Table of Contents

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. This report presents a literature review of the latest developments in the computational modeling area of neck biomechanics. Mature computer models of the head and neck are reviewed to determine the most biofidelic and versatile model available. Also, full body computational models of the human are evaluated with specific emphasis on the neck region. The intent of this report is to provide a detailed review of the available models so that an appropriate model can be selected to obtaining the necessary data for the specific application of studying carotid artery injuries to vehicle occupants in far-side impacts. It should be noted that this review can only focus on publicly available information regarding computational models of the neck, there is the possibility that other proprietary or other similar models of the neck are used for biomechanical modeling; however, those models are not included as they are not discussed in the open literature.

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. However, computational models are useless 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 parallel 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 muscle improvements such as the addition of the proper mass and more detailed geometry should be the focus of 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 1). 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). Because it is a two-dimensional model, it is not of use for studying lateral impacts.

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 intervertebral-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. 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 (Bosio 1986, Bowman 1972, Bowman 1975, Bowman 1981, Bowman 1984, Goldsmith 1984, Li 1991, McElhaney 1979, Melvin 1972, Nightingale 2000, Paver 1990, Seemann 1984, Tien 1985, Tien 1987).

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. At the time, run times were 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 (Lizee 1998). The FE neck was then integrated into a 50th percentile seated male FE 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.

A 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 has 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 FE 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 (Frechede 2003, Roychoudhury 2000, Yang 1992, Yoshida 2002).

2.4. Multibody-Finite Element Combination Models

Perhaps the most effective method to study a particular region on a large scale model, multibody-FE combination models allow the user to model the area of interest with FE 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.

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 FE head with rigid body vertebra. Like many of its computational predecessors, this model has no neck musculature. According to the authors, this is because musculature is not thought to be able to react fast enough to affect the response of the neck under compressive impacts (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 completely on material properties to determine accurate three-dimensional kinematics. Also, they state that material-based tolerance criteria lack injury correlation with predicted stresses and strains.

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

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 (1996a, 1996b). 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 of 3D global mesh geometry, 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. 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 FE 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). Also, the computer model anatomy is detailed enough, despite the lack of FE deformation, 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), 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 both PMHS and 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.

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, it is imperative that proper positioning of the occupant 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-FE 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 can offer biofidelic responses on par with current FE element models but without details of local stresses and strains. Additionally, multibody-FE 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. Specifically concerning detailed neck models, it is currently the model of choice to use in a computational simulation of neck response to dynamic loading conditions.

3. Full Body Human Models

3.1. Introduction to Full Body Models

For the purposes of studying carotid artery injuries to far-side occupants, it is necessary to not only have a detailed model of the neck, but also to have a full body model of the occupant. These full body models will allow the researcher to study the interaction of the neck with the shoulder belt and kinematics of the head, which alone or in combination are commonly thought to be mechanisms for injury for the carotid artery. Many of the models discussed include detailed models of the neck from the previous section; however, the purpose of this portion of the paper is to provide an overview on the different full body models that have been presented in the recent literature.

3.2. Multibody Models

Although there are many full body multibody models of the human occupant available in the literature, their usefulness is limited, given the availability of FE and Multibody-FE combo models with detailed models of the neck. Therefore, no multibody ellipsoid models were evaluated for their usefulness for this project.

3.3. Finite Element Models

There are several projects underway to develop a useful FE model of the whole human body. The advantage to this technique is that it provides a platform that can be theoretically used in a multitude of crash or impact scenarios, if the model is carefully constructed to offer accurate replication of the structure of the human body.

As mentioned previously, the component FE model of the neck developed by Dauvilliers et al. (1994) was integrated into a seated occupant model by Lizee et al. (1998). This full body model was intentionally kept geometrically simple, as it was intended for use in full vehicle models and it was not desired that the addition of the occupant would adversely affect the run time of the simulation. The model size was limited to 10,000 elements; however, even with its inherent simplicity the model was able to accurately predict occupant responses to frontal, lateral, and some rear impact scenarios. The model does not have enough detail, in current form, to predict injury to specific regions of the body; however, with detailed improvement it should not only be able to predict accurate kinematics but also localized loads of tissues.

A model referred to as the H-Dummy was presented by Choi et al. (1999). The model is intended to provide an omni-directional model that can be used in a variety of impact situations. The model contains a detailed cervical spine with vertebrae, ligaments, discs and muscles. Geometric data incorporated into the model was based on Viewpoint Datalabs' dataset and various anthropometric textbooks. No discussion is given to the selection of material properties or the active/passive response of the muscle models. The cervical spine motion was validated both to local response of each pair of adjacent vertebrae and global response of the neck, to previously published data. Additional data was published by Choi et al. (2002) validating the response of the model specifically to low speed rear impacts.

Another FE model of the human occupant is the Total Human Model for Safety (THUMS) (Iwamoto 2002). This model presents a FE model of the bones, ligaments, tendons, flesh, and skin, there are also detailed models of the shoulder, head and face, or internal organs, if those are of interest for the simulation. As in the H-Dummy, geometric data was taken from Viewpoint Datalabs and anatomy textbooks. The neck muscles can be flexed to simulate active musculature; however, they are tension only members utilizing bar elements. The thorax and spine segments of the model have been validated for side impacts and are reported to fall within published cadaver response corridors. With approximately 83,500 elements, the base model runs with a minimal amount of CPU time and is able to simulate detailed injuries to the cervical spine and thorax.

Ford Motor Company has been developing a FE human body model that operates in the PAM-CRASH environment (Ruan 2003). The model consists of approximately 103,000 nodes and 119,000 elements. Geometry of the model is based on the Visible Human Project and anatomical texts and incorporates an updated version of the neck model previously developed by Nitsche et al. (1996). Originally validated by Nitsche, the model has been further elaborated with new facet joints along with new and remodeled ligaments. At present, the integrated neck model has not been validated in lateral impacts, although in component testing, the original neck model (Nitsche 1996) was validated to frontal, lateral and rear impact scenarios. As discussed previously, in those validation tests the head-neck motion was compared to that from frontal and lateral volunteer tests and displayed better correlation with frontal test data than with lateral test data. It is unknown whether the improvements the model have improved the lateral impact response. The integrated full body model has been validated to lateral thoracic impact.

Currently, such an effort is underway at Wayne State University, to develop an advanced FE model of the human (Yang 2003). The model is a combination of human component models that have been developed based on the scale of a 50th percentile male. This model incorporates the detailed neck model presented by Yang et al. (1998), which was discussed previously. Geometric data for the neck model comes from an MRI of an approximate 50th percentile male individual, other geometry is taken from the Visible Human Project. The neck model incorporates detailed geometry of the neck, passive muscle bar elements, and intervertebral discs; however, the model does not yet have the capability for active muscle simulation. The fully developed and integrated model is expected to be completed sometime in the next five years.

Lastly, a human body model consisting of the THUMS and Engineering Systems International's H-Dummy (Murakami 2004). The upper body of this integrated model is taken from THUMS, while the lower extremities of the model are developed from the H-model. This model contains approximately 67,000 nodes and the geometry of the vertebrae are accurately modeled using rigid body solid elements. As this model's thorax, head, and neck originated from the THUMS model, it can be assumed that geometry originated from Viewpoint Datalabs and anatomical text, although details of further geometrical refinement are not offered. Intervertebral discs are modeled as deformable solid elements and cervical ligaments are modeled as membrane elements. Muscles consist of active and passive Hill-type elements. Neck response of this model was compared to a volunteer whiplash simulation where good correlation was seen in the initial phase of motion, until maximum extension; however, the motion after the maximum extension did not correlate with the human volunteer responses. Further work is necessary to optimize the head-neck responses of this model to impact scenarios, with possible improvements coming in the form of soft tissues of the neck and improved muscle element modeling.

3.4. Multibody-Finite Element Combination Models

As mentioned previously, a full body human model with a detailed neck exists in the MADYMO software environment (Happee 1999, Van der Horst 2002). This model contains the detailed neck model developed by Van der Horst (2002) which is a rigid-body model utilizing detailed geometry of facet joints and ligaments, as well as anatomically accurate, curved neck muscles. Geometry was defined by dissection of a frozen cadaver in a seated automotive posture. No FE techniques are currently employed in the MADYMO full body model with integrated detailed neck. MADYMO is capable of performing FE simulations and a FE human model does exist; however, the geometry of the cervical spine is less detailed in the FE human. Also, as mentioned previously, lateral impact validation has been performed using the MADYMO detailed neck model (Meijer 2003). Although validation tests were compared to a small number of physical experiments, the model showed good overall correlation to the limited number of tests.

3.5. Full Body Model Summary

The ability to duplicate the positioning of the occupant in the vehicle and replicate the total response of the occupant in an impact, including interaction with the restraint system is the main advantage the full body models offer over the component neck models. When using a component model, it is necessary to understand the orientation and constraints of the whole body so that the component model can be setup in a life-like scenario; however, the use of a full body model allows a more intuitive model setup, with an occupant's position and restraint systems mimicking the physical setup. In the last several years, the level of detail in full body models has begun to approach that originally seen only in component level models. As a result, the use of a full body models to gain information on loads and kinematic responses seen by the occupant is becoming more and more prevalent. The individual full body models reviewed in this chapter will certainly help the researcher understand the biomechanics of impact to a greater level of detail then ever before; however, each model presents its own advantages and disadvantages (Table 5).

Table 5 – Summary Table of Full Body Models

Author (Year)	Program	Number of Elements	Loading Conditions[1]	Advantages	Disadvantages
Lizee (1998)	RADIOSS	10,000 elements	Frontal impact; Lateral impact; Rear impact	Computationally efficient; Good global validation	Rudimentary geometry; No neck musculature; Lack of detail for interpretation on local tissue level; No soft tissue modeling around neck
Choi (1999)	PAM-CRASH	Unknown	Rear impact	Some soft tissue modeling of vertebrae, muscles, ligaments, organs, discs, and fatty tissues in neck and torso	Only validated to low speed rear impacts; Unknown level of detail of soft tissue in neck
Iwamoto (2002)	PAM-CRASH/ LS-DYNA	83,500 elements	Frontal impact; Lateral impact; Rear impact	Thorax and spine segments of model validated for lateral impacts; Active neck musculature	Lacks overall validation of complete model; No soft tissue modeling around neck
Ruan (2003)	PAM-CRASH	103,000 nodes; 119,000 elements	Frontal impact; Lateral impact; Rear impact	Detailed neck model with new and remodeled facet joints and ligaments	Lacks overall validation of model; No neck musculature; No soft tissue modeling around neck
Yang (2003)	PAM-CRASH	Unknown	Rear impact; Frontal impact	Marriage of separate detailed component models (head, neck, thorax, abdomen, lower limb)	Incomplete; No active musculature; No soft tissue modeling around neck
Murakami (2004)	UNKNOWN	67,000 nodes	Rear impact	Optimized mesh quality and calculation time (over THUMS and H-Dummy contributors); Active neck muscles	Low level validation; No soft tissue modeling around neck
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 neck muscles; Detailed for local and global injury assessment	FE techniques would offer better local injury assessment; Lack of high severity muscle activation data; No soft tissue modeling around neck

[1] All of these models are intended for ultimate use in any loading direction. Loading conditions presented here are for impact directions that neck component has been subjected to.

These models presented are all in various states of development and have undergone various states of validation. For the specific purposes of studying far-side impacts, the best choice of models from those available would be one that has had extensive development and validation to lateral impacts. With that in mind, several models should likely be omitted from lateral impact studies in their current form. Although no models have provided extensive validation data for lateral impacts, the Iwamoto et al. (2002) and Van der Horst (2002) models have seemingly had the most development for lateral impact scenarios. The Van der Horst model has been globally validated to a limited number of lateral scenarios and has displayed reasonable agreement, although to a small sample size of validation tests. The Iwamoto model has not been globally validated to lateral impacts; however, the thorax and spine segments of the model have been validated to lateral impacts. Since these are likely the most important areas of the body for studying carotid artery injury, it is felt that this model will have sufficient biofidelity in its current state for these types of simulations.

4. Conclusion

Despite the obvious complexities of the human body, it has been shown that a myriad of computational models of the neck exist. The level of detail in the models presented in the literature varies widely, from low level detail offering rudimentary kinematic correlation, to highly detailed models with accurate geometry and simulation of active muscle control. For the purposes of studying the effects of injuries to the carotid artery for occupants on the far-side of the struck vehicle, it is necessary to have a detailed understanding of the kinematics of the neck under lateral impact and to understand the local loading mechanisms of the tissues of the neck, preferably with a detailed understanding of geometrical interactions. It is difficult to make a judgment on which model has the most accurate anatomical data, but it can be said that the nodal makeup and mesh size of even the most finely meshed model in this review is quite large compared to what would be required for accurate modeling of the tissue of the carotid artery itself. Considering that a co-requirement to accurate geometry is that the model is validated for lateral impacts, the two most possibly useful models are the THUMS model by Iwamoto et a. (2002) and the MADYMO model by Van der Horst (2002). Both of these models will be able to provide the most accurate representation both globally and locally of the kinematics of the neck under lateral bending; however, both of these models are limited in that they do not fully represent the soft tissues around the neck, rather incorporate major tissues such as the bony structures, discs, and ligaments, while representing the muscle lines of action. Aside from a shroud of skin mesh around the models, no localized interactions with soft tissues such as fat or perturbation of the muscles are accounted for. However, these models contain sufficient features that can be monitored during simulations to extract meaningful data for validation of a more specialized local model of the cervical neck with accurate soft tissue representations.

References

Bomar, J.B., and Pancratz, D.J. (1998) A PC-Based Head-Spine Model. NATO RTO Meeting Proceedings 20; Models for Aircrew Safety Assessment: Uses, Limitations and Requirements: 4-1-4-11 North Atlantic Treaty Organization Research and Technology Organization.

Bosio, A.C., and Bowman, B.M. (1986) Simulation of Head-Neck Dynamic Response in –Gx and +Gy. Stapp Car Crash Journal 30: 345-378. SAE 861895.

Bowman, B.M., and Robbins, D.H. (1972) Parameter Study of Biomechanical Quantities in Analytical Neck Models. Stapp Car Crash Journal 16: 14-44. SAE 720957.

Bowman, B.M., and Schneider, L.W. (1975) Simulated Occupant Response to Side-Impact Collisions. Stapp Car Crash Journal 19: 429-454. SAE 751155.

Bowman, B.M., Schneider, L.W., Rohr, P.R., and Mohan, D. (1981) Simulation of Head/Neck Impact Responses for Helmeted and Unhelmeted Motorcyclists. Stapp Car Crash Journal 25: 13-68. SAE 811029.

Bowman, B.M., Schneider, L.W., Lustick, L.S., Anderson, W.R., and Thomas, D.J. (1984) Simulation Analysis of Head and Neck Dynamic Response. Stapp Car Crash Journal 28: 173-205. SAE 841668.

Brelin-Fornari, J.M. (1998) A Lumped Parameter Model of the Human Head and Neck with Active Muscles. PhD thesis, University of Arizona, USA.

Camacho, D.L., Nightingale, R.W., Robinette, J.J., Vanguri, S.K., Coates, D.J., and Myers, B.S. (1997) Experimental Flexibility Measurements for the Development of a Computational Head-Neck Model Validated for Near-Vertex Head Impact. Stapp Car Crash Journal 41: 473-486. SAE 973345.

Chancey, V.C., Nightingale, R.W., VanEe, C.A., Knaub, K.E., and Myers, B.S. (2003) Improved Estimation of Human Neck Tensile Tolerance: Reducing the Range of Reported Tolerance Using Anthropometrically Correct Muscles and Optimized Physiologic Initial Conditions. Stapp Car Crash Journal 47. SAE 2003-22-0008.

Choi, H.Y., Eom, H.W., Kho, S.T., Lee, I.H. (1999) Finite Element Human Model for Crashworthiness Simulation. SAE 1999-01-1906.

Choi, H.Y., Lee, I.H., Haug, E., and Lee, J.H. (2002) Finite Element Human Model for Low Speed Rear Impact Situation. Proc. of the 2002 World Congress of Biomechanics.

Dauvilliers, F., Bendjellal, F., Weiss, M., Lavaste, F., and Tarriere, C. (1994) Development of a Finite Element Model of the Neck. Stapp Car Crash Journal 38: 77-91. SAE 942210.

Deng, Y.C., and Goldsmith, W. (1987) Response of a Human Head/ Neck/ Upper-Torso Replica to Dynamic Loading – II. Analytical/ Numerical Model. Journal of Biomechanics 20: 487-497.

Deng, Y.C., Li, X., Liu, Y. (1999) Modeling of the Human Cervical Spine Using Finite Element Techniques. SAE 1999-01-1310.

Deng, Y.C., Fu, J. (2002) Simulation and Identification of the Neck Muscle Activities During Head and Neck Flexion Whiplash. SAE 2002-01-0017.

De Jager, M., Sauren, A., Thunnissen, J., and Wismans, J. (1994) A Three-Dimensional Head-Neck Model: Validation for Frontal and Lateral Impacts. Stapp Car Crash Journal 38: 93-109. SAE Paper Number 942211.

De Jager, M., Sauren, A., Thunnissen, J., and Wismans, J. (1996a) A Global and a Detailed Mathematical Model for Head-neck Dynamics. Stapp Car Crash Journal 40: 269-281. SAE Paper 962430.

De Jager, M.K.J. (1996b) Mathematical Head-Neck Models for Acceleration Impacts. PhD thesis, University of Eindhoven.

Frechede, B., Saillant, G., Lavaste, F., and Skalli, W. (2003) Relationship Between Cervical Spine Curvature and Risk of Injury in the Case of Sagittal Impact: A Finite Element Analysis. 2003 International IRCOBI Conference on the Biomechanics of Impacts: 355-356.

Goldsmith, W., Deng, Y.C., and Merrill, T.H. (1984) Numerical Evaluation of the Three-Dimensional Response of a Human Head-Neck Model to Dynamic Loading. In SAE Government Industry Meeting and Exposition. SAE 840861.

Halldin, P.H., Brolin, K., Kleiven, S., Von Holst, H., Jakobsson, L., Palmertz, C. (2000) Investigation of Conditions That Affect Neck Compression-Flexion Injuries Using Numerical Techniques. Stapp Car Crash Journal 44. SAE 2000-01-SC10.

Happee, R., Morsink, P., and Wismans, J. (1999) Mathematical Human Body Modeling for Impact Loading. In Digital Human Modeling for Design and Engineering, International Conference The Hague: 1-9. SAE 1999-01-1909.

Happee, R., Ridella, S., Nayef, A., Morsink, P., de Lange, R., Bours, R., and Van Hoof, J. (2000) Mathematical Human Body Models Representing a Mid Size Male and a Small Female For Frontal, Lateral and Rearward Impact Loading. 2000 International IRCOBI Conference on the Biomechanics of Impacts.

Hayamizu, N., Miki, K., and King, A.I. (1999) Analysis of Head and Neck Response During Side Impact. In SAE International Congress and Exposition. SAE 1999-01-0717.

Huston, R.L., Huston, J.C., and Harlow, M.W. (1978a) Comprehensive, Three-Dimensional Head-Neck Model for Impact and High-Acceleration Studies. Aviation, Space, and Environmental Medicine 49: 205-210.

Huston, R.L., and Passerello, C.E. (1978b) Multibody Dynamics Including Translation Between the Bodies – With Application to Head-Neck Systems. Report Number AD-A062114.

Iwamoto, M., Kisanuki, Y., Watanabe, I., Furusu, K., Miki, K., Hasegawa, J. (2002) Development of a Finite Element Model of the Total Human Model for Safety (THUMS) and Application to Injury Reconstruction. 2002 International IRCOBI Conference on the Biomechanics of Impacts: 31-42.

Jakobsson, L., Norin, H., Jernstrom, C., Svensson, S.E., Johnsen, P., Isaksson-Hellman, I., and Svensson, M.Y. (1994) Analysis of Different Head and Neck Responses in Rear-End Car Collisions Using a New Humanlike Mathematical Model. 1994 International IRCOBI Conference on the Biomechanics of Impacts: 109-125.

Kleinberger, M. (1993) Application of Finite Element Techniques to the Study of Cervical Spine Mechanics. Stapp Car Crash Journal 37: 261-272.

Li, Y., Bishop, P.J., Wells, R.P., and McGill, S.M. (1991) A Quasi-Static Analytical Sagittal Plane Model of the Cervical Spine in Extension and Compression. Stapp Car Crash Journal 35: 419-433. SAE 912917.

Lizee, E., Robin, S., Song, E., Bertholon, N., Le Coz, J.Y., Besnault, B., and Lavaste, F. (1998) Development of a 3D Finite Element Model of the Human Body. Stapp Car Crash Journal 42: 115-138. SAE 983152.

McElhaney, J., Snyder, R.G., States, J.D., and Gabrielsen, M.A. (1979) Biomechanical Analysis of Swimming Pool Neck Injuries. Automotive Engineering Congress and Exposition. SAE 790137.

Melvin, J.W., McElhaney, J.H., and Roberts, V.L. (1972) Improved Neck Simulation for Anthropometric Dummies. Stapp Car Crash Journal 16: 45-60. SAE 720958.

Meijer, R., Parenteau, C., Van Hoof, J., and Gopal, M. (2003) Validation of a MADYMO Mathematical Human Body Model with Detailed Neck in Low Speed Lateral Impacts. 2003 International IRCOBI Conference on the Biomechanics of Impacts: 357-358.

Murakami, D., Kitagawa, Y., Kobayashi, S., Kent, R., and Crandall, J. (2004) Development and Validation of a Finite Element Model of a Vehicle Occupant. SAE 2004-01-0325.

Nightingale, R.W., Camacho, D.L., Armstrong, A.J., Robinette, J.J., and Myers, B.S. (2000) Inertial Properties and Loading Rates Affect Buckling Modes and Injury Mechanisms in the Cervical Spine. Journal of Biomechanics 33: 191-197.

Nitsche, S., Krabbel, G., Appel, H., and Haug, E. (1996) Validation of a Finite Element Model of the Human Neck. 1996 International IRCOBI Conference on the Biomechanics of Impacts: 107-122.

Paver, J.G., Khatha, T.P., Piziali, R.L., Whitestone, J., Kaleps, I., and Taylor, C. (1990) The Prediction of Hybrid III Manikin Head-Neck Kinematics and Dynamics. SAE International Congress and Exposition. SAE 900540.

Roychoudhury, R., Sun, D., Hamid, M., and Hanson, C. (2000) 5th Percentile Driver Out of Position Computer Simulation. SAE 2000 World Congress. SAE 2000-01-1006.

Ruan. J., El-Jawahri, R., Chai, L., Barbar, S., Prasad, P. (2003) Prediction and Analysis of Human Thoracic Impact Responses and Injuries in Cadaver Impacts Using a Full Human Body Finite Element Model. Stapp Car Crash Journal 47. SAE paper number 2003-22-0014.

Seemann, M.R., Lustick, L.S., and Frisch, G.D. (1984) Mechanism for Control of Head and Neck Dynamic Response. Stapp Car Crash Journal 28: 207-222. SAE 841669.

Tien, C.S., and Huston, R.L. (1985) Biodynamic Modelling of the Head/Neck System. SAE International Congress and Exposition. SAE 850438.

Tien, C.S., and Huston, R.L. (1987) Numerical Advances in Gross-Motion Simulations of Head/Neck Dynamics. Journal of Biomechanical Engineering, 109: 163-168.

Van den Kroonenberg, A., Thunnissen, J., and Wismans, J. (1997) A Human Model for Low-Severity Rear-Impacts. 1997 International IRCOBI Conference on the Biomechanics of Impacts: 117-132.

Van der Horst, M.J., Thunnissen, J.G.M., Happee, R., Van Haaster, R.M.H.P., and Wismans, J.S.H.M. (1997) The Influence of Muscle Activity on Head-Neck Response During Impact. Stapp Car Crash Journal 41: 487-507. SAE 973346.

Van der Horst, M.J., Bovendeerd, P.H.M., Happee, R., Wismans, J.S.H.W., and Kingma, H. (2001) Simulation of Rear End Impact with a Full Body Human Model with a Detailed Neck: Role of Passive Muscle Properties and Initial Seating Posture. Proc. ESV Conference 17, Amsterdam, The Netherlands. NHTSA, USA. Paper Number 119.

Van der Horst, M.J. (2002) Human Head Neck Response in Frontal, Lateral and Rear End Impact Loading: Modeling and Validation. The Netherlands: PhD dissertation, University of Eindhoven.

Van Ee, C.A., Chasse, A.L., and Myers, B.S. (2000) Quantifying Skeletal Muscle Properties in Cadaveric Test Specimens: Effects of Mechanical Loading, Postmortem Time, and Freezer Storage. Journal of Biomechanical Engineering, 122: 9-14.

Van Ee, C.A., Nightingale, R.W., Camacho, D.L.A., Chancey, V.C., Knaub, K.E., Sun, E.A., and Myers, B.S. (2000) Tensile Properties of the Human Muscular and Ligamentous Cervical Spine. Stapp Car Crash Journal 44. SAE 2000-01-SC07.

Van Hoof, J., Meijer, R., and Van der Horst, M. (2002) Effects of Muscle Activation and Seating Posture in Rear End Impact Using a Human Model with a Detailed Neck. Proc. 2002 World Congress of Biomechanics.

Weerappuli, D.P.V., Sin, D.Z., and Stanecki, P. (1998) Predicting the Response of a CART Driver in a Crash Using Mathematical Modeling. In SAE Motorsports Engineering Conference and Exposition. SAE 983069.

Williams, J.L., and Belytschko, T.B. (1983) A Three-Dimensional Model of the Human Cervical Spine for Impact Simulation. Journal of Biomechanical Engineering, 105: 321-331.

Winters, J.M. (1990a) Hill-Based Muscle Models: A Systems Engineering Perspective. Winters, J.M., and Woo, S.L.Y. Multiple Muscle Systems: Biomechanics and Movement Organization: 69-93. Springer-Verlag.

Wismans, J.S.H.M., Van den Kroonenberg, A.J., Hoofman, M.L.C., and Van der Horst, M.J. (1998) Neck Performance of Human Substitutes in Frontal Impact Direction. In NATO RTO Meeting Proceedings 20; Models for Aircrew Safety Assessment: Uses, Limitations and Requirements: 5-1-5-6. North Atlantic Treaty Organization Research and Technology Organization.

Yamazaki, K., Ono, K., and Kaneoka, K. (2000) A Simulation Analysis of Human Cervical Spine Motion During Low Speed Rear-End Impacts. SAE International Congress and Exposition. SAE 2000-01-0154.

Yang, K.H., and Le, J. (1992) Finite Element Modeling of Hybrid III Head-Neck Complex. Stapp Car Crash Journal 36: 219-233. SAE 922526.

Yang, K.Y., Zhu, F., Luan, F., Zhoa, L., and Begeman, P.C. (1998) Development of a Finite Element Model of the Human Neck. Stapp Car Crash Journal 42: 195-205. SAE 983157.

Yang, K.H., Beillas, P., Zhang, L., Lee, J.B., Shah, C., Hardy, W., Demetropoulos, C., Tashman, S., and King, A.I. (2003) Advanced Human Modeling for Injury Biomechanics Research. SAE 2003-01-2223.

Yoshida, H., and Tsutsumi, S. (2002) Finite Element Analysis Using Muscle Elements and Experimental Analysis With a New Flexible Neck Model of Whiplash Injuries in Rear-End Collisions. SAE 2002-01-0021.