AIMING FOR AN AVERAGE FEMALE VIRTUAL HUMAN BODY MODEL FOR SEAT PERFORMANCE ASSESSMENT IN REAR-END IMPACTS

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ABSTRACT

The female part of the population suffers more Whiplash Associated Disorders (WAD) in car crashes than males. Several studies have illustrated the need to consider the female population when developing and assessing the WAD prevention performance of advanced restraint systems in rear-end collisions. Presently only one crash test dummy is available, the average sized male BioRID. Recently a virtual dummy model of an average female, EvaRID, was developed and used in rear impact simulations. The results stressed the need for models representing the female part of the population, as well. Virtual crash simulations have become essential in traffic safety and with models of both an average male and female, further steps in addressing improved assessment of WAD prevention can be taken. The present paper presents a starting point of research aiming to develop an open-source average female Finite Element (FE) model with an anatomically detailed cervical spine. This paper provides a review of the literature to identify gender specific neck biomechanics and anatomical differences, followed by a review of published FE models of the cervical spine.

Data on vertebral body dimensions (height, width, depth, spinal canal diameter, facet joint angles) have been compiled from biomechanical literature. Significant gender differences exist for the vertebral body depth and width, the spinal curvature in the seated posture, and the spinal stiffness and range of motion. All have the potential to influence the outcome of an impact and should be accounted for in the development of WAD prevention.

The review of FE models of the cervical spine presented 17 models based on male geometry but only one model scaled to represent a female. An overview of the models are given with respect to the solver, geometry source, number of elements, and implementation of the facet joints, ligaments, and muscles. It is recommended that an average female model is developed with focus on; 1) the shape of the female vertebral body, especially the depth and width that provides less support area than for males, 2) defining the spinal curvature representative of seated female volunteers who generally display less lordosis than males, 3) the dimensions of the spinal ligaments, rather than the material properties, to capture the larger range of motion and less spinal stiffness of female subjects compared to males, and 4) validation to female volunteers and PMHS tests for range of motion, while failure prediction seem less gender sensitive.

KEYWORDS**:** Whiplash Associated Disorders, gender, biomechanics, finite element human body model

INTRODUCTION

Whiplash Associated Disorders (WAD) is the common denomination for a range of neck related symptoms that typically occur in vehicle collisions. During a rear-end collison, the torso of a properly restrained occupant will be accelerated while the head tends to lag behind causing inertial loading to the neck. This loading causes a so called whiplash motion of the neck in which the neck is forced into an unphysiological s-curved shape that later evolves into a c-shape (Ono et al, 1997). For example, in Sweden WADs account for ~70% of all injuries leading to disability due to vehicle crashes (Kullgren et al. 2007). The majority of victims experiencing initial WAD symptoms recover within a few weeks or months of the crash (The Whiplash Commission 2005), however, 5–10% of individuals experience different levels of medically classified permanent disabilities (Nygren 1983, Krafft 1998, The Whiplash Commission 2005). WADs occur at relatively low velocity changes, typically <25 km/h (Eichberger et al. 1996, Kullgren et al. 2003), and in impacts from all directions, although rear impacts are most frequently featured in accident statistics (Watanabe et al. 2000). Since the mid-1960s, injury statistical data have shown that females have a higher risk of sustaining WAD than males, even in similar crash conditions (Narragon 1965, Kihlberg 1969, O'Neill et al. 1972, Thomas et al. 1982, Otremski et al. 1989, Maag et al. 1990, Morris & Thomas 1996, Dolinis 1997, Temming & Zobel 1998, Richter et al. 2000, Chapline et al. 2000, Krafft et al. 2003, Jakobsson et al. 2004, Storvik et al. 2009, Carstensen et al. 2012). According to these studies, WAD risk is 1.4 to 3 times higher for females compared to males.

Currently rear impact tests are performed using the only available dummy developed for low severity testing, the Biofidelic Rear Impact Dummy, BioRID II (Davidsson et al. 1999), which represents a 50th percentile male. The BioRID II is used in test protocols, i.e. Euro NCAP, for assessing the risk of whiplash injuries when seated in the front seat. The BioRID was developed in the late 1990s to evaluate the protective performances of car seats in low severity rear impacts. However, real world accident analysis has shown that seats designed to prevent WAD are more effective for males (Kullgren et al., 2013). Therefore, there is an urgent need to better represent the whole adult population in the development and assessment of WAD prevention systems and develop models representing females, as well as males for assessing the prevention performance.

The initial step in the work towards addressing WADs in females was recently taken in the ADSEAT project (Linder et al. 2013) that adopted a broad approach when developing the world first average female virtual dummy model, EvaRID (Carlsson et al. 2014). Virtual impact simulations with seats showed, using the virtual male and female dummy models, that differences were found in the response of the BioRID II and EvaRID models for some of the seat concepts tested (Linder et al 2013). Furthermore, a prototype dummy of an average female was developed and run in the same test set-up as the Euro NCAP tests performed with the BioRID II (Schmitt et al. 2012). Four different seat types were tested and the results confirmed that the size of males and females interact differently with the seat and the head restraint. Based on the dummy responses, male and female occupants would thus obtain different levels of protection when seated in the same seat. Hence, having access to models of both an average male and female, further steps in addressing improved assessment of WAD prevention can be taken. Virtual models offer a greater variety of seated postures and the potential to evaluate a wider range of parameters, compared to what is practically possible in physical testing.

Virtual crash simulations using the Finite Element (FE) method has become an essential tool in traffic safety and with increasing computer capacity the size and complexity of the models has increased. Traditionally, FE occupant models were virtual copies of crash test dummies but more recently there has been a strong progression in the development of Human Body Models (HBMs). The latter have been designed to represent the human body rather than a dummy and have the potential to allow for much more detailed studies of injury mechanisms than physical or virtual dummies. A number of occupant HBMs have been released, primarily of the average male anthropometry, but are not widely used due to a number of reasons such as model quality, restricted availability or high licencing fees. Therefore, the need for open source models where knowledge and experience can be shared and continued improvements of the models can be provided by any user is apparent. Our overall aim is to develop an open-source FE model of an average female, with an anatomically detailed cervical spine suitable for assessment of seat performance.

This paper provides a review of the literature to identify gender specific neck biomechanics and anatomical differences followed by a review of published FE models of the cervical spine. Lastly, a short complementary discussion presents the conlusion on requirements for the development of an average female HBM. The resulting

recommendations have the potential to improve the biofidelity of female HBMs and thereby enhance the development and assessment of safety systems that can reduce the incidence of WAD in traffic accidents.

INJURY MECHANISMS

Despite much research, WAD is still one of the most poorly understood traffic injuries. Literature supports an organic basis for WADs (Siegmund et al. 2009; Curatolo et al. 2011). Several anatomical sites have been proposed, including, facet joints, spinal ligaments, intervertebral discs, vertebral arteries, dorsal root ganglia, and neck muscles. Each of these tissues is strained during a whiplash motion exposure. Two mechanisms of facet joint lesions have been proposed: pinching of the synovial fold and excessive capsule strain. The cervical vertebrae have been shown to rotate about an elevated instantaneous centre during whiplash motion in a rear-end collision. This may compress the posterior facet surfaces, pinching the synovial fold (Ono et al. 1997; Kaneoka et al. 1999). Excessive facet capsule strain has been demonstrated during rear-end collision whiplash motion experiments where, in a functional cervical spinal unit, the upper vertebrae was exposed to excessive rearward shear (Pearson et al. 2004; Yang and King 2003). Lesions to the other neck ligaments and intervertebral discs have been reported by Krakenes and Kaale (2006), for instance. Ligament lesions may cause acute neck pain and lead to chronic spinal instability. Lesion to mechanoreceptors may corrupt the normal sensory signals and can lead to abnormal muscle response patterns, decreased neck mobility and proprioception (Panjabi et al. 2006).

Chronic symptoms such as headache, blurred vision, tinnitus, dizziness, and vertigo have been proposed to be associated with altered blood flow rates due to spasm and/or narrowing of vertebral arteries in WAD patients (Reddy et al. 2002; Seric et al. 2000). Intimal tears of the vertebral artery are most common at the atlanto-axial joint. This has been hypothesized to be caused by coupled extension and axial rotation of the upper cervical spine. Also, many WAD symptoms could be explained by lesion to the dorsal root ganglia that contain the cell bodies of most peripheral sensory nerves, at each spinal level. Increased electrical activity in the spinal cord and widespread reductions in electrical and pressure thresholds after whiplash motion exposure suggest altered central pain processing (Banic et al. 2004, Curatolo et al. 2001, Kasch et al. 2001a, Scott et al. 2005). During whiplash motion, pressure transients have been registered in the spinal canal (Örtengren et al. 1996, Svensson et al. 2000). These are hypothesized to be the cause of spinal ganglion nerve cell membrane dysfunction. Direct deformation of the nerve roots is another possible cause as the diameter of the neural foramina decrease during extreme neck motions.

Muscle pain is a common WAD symptom, although evidence of direct lesions in muscles remains inconclusive. Direct muscle lesions may not be responsible for chronic pain, but may play an indirect role in modulating pain caused by injuries to other structures. The direct mechanism of muscle lesions occur due to imposed lengthening during active contraction. Both anterior and posterior muscles experience active lengthening (Brault et al. 2000; Vasavada et al. 2007). Neck muscles interact with other anatomical sites: 1) they attach to the facet capsule; 2) they indirectly load other neck structures; and 3) altered neuromuscular control may contribute to chronic pain.

For each of the above mentioned tissues, continued research is needed to help improve diagnosis and treatment. Simulations with HBMs can help shed light on the strain in different neck tissues during the multitude of different loading scenarios that occur in traffic accidents and are known to produce WADs. Also, in order to assess the protective properties of vehicle seats and other safety systems it is necessary for the HBMs to be detailed enough to provide metrics for all or at least most of the above mentioned injury mechanisms. To predict tears of the vertebral arteries and excessive strain of upper cervical ligaments, it is essential to model the complex structure of the upper cervical joints with sliding contacts between the joints and soft tissues limiting the motion. Facet joint motion and strain in the spinal ligaments are highly dependent on the vertebral kinematics and therefore it is essential the HBMs have biofidelic relative motion in the vertebral segments.

ANTHROPOMETRY AND MATERIAL PROPERTIES

There are significant differences in neck anthropometry based on gender. The female neck is more slender than the male neck, for example Harty et al. (2004) found the female neck circumference to be 84% of the male neck circumference and female neck length was 91% of the male. There is little gender differences in the head circumference while the neck circumference is smaller for females (Vasavada et al. 2001, Harty et al. 2004,

Mordaka 2004, Valkeinen et al. 2002, Vasavada et al. 2008, DeRosia 2008). Hence, females support more head mass per neck cross-sectional area. These gender differences may to some extent depend on stature and not directly on gender. To distinguish the effects of gender from stature, size matched male and female subjects were studied by Vasavada et al. (2008) and DeRosia (2008) (Table 1). It can be seen that head circumference varied by only 3%, irrespective of size matching based on neck length, seated height, or stature. On the other hand, the female neck circumference was 84 – 86 % of the male measurements for all groups. Hence, gender have significant effect on neck circumference, which is not an effect of differences in overall body dimensions. Gender differences for neck length were less than the difference for neck circumference, supporting previous findings that female necks are more slender (Vasavada et al. 2008, DeRosia 2008).

Table 1. Body Dimensions for Size Matched Volunteers

F = Females, M = Males, No. = Number of subjects, Circumf. = Circumference.

*Neck length measure: C7 spinous process - tragus of the ear (Vasavada 2008), C2-C6 length (DeRosia 2008).

		average ± SD				
References		Schneider et al. 1983	Vasavada et al. 2008	DeRosia 2008	DeRosia 2008	DeRosia 2008
Size Matched		50th percentiles	neck length	seated height	head circumf.	stature
No. F			14	11	19	11
No. M			14	17	15	10
Head Circumf.	F		562 ± 15	561 ± 14	559 ± 7	548 ± 15
[mm]	M		577 ± 13	580 ± 15	561 ± 8	564 ± 21
	F/M		97%	97%	100%	97%
Neck Circumf.	F		331 ± 20	324 ± 12	322 ± 14	318 ± 22
[mm]	M		394 ± 22	376 ± 16	379 ± 20	379 ± 23
	F/M		84%	86%	85%	84%
Neck length*	F		107 ± 5	88 ± 6	86 ± 6	86 ± 7
[mm]	M		108 ± 5	95 ± 6	97 ± 6	97 ± 6
	F/M		99%	94%	88%	89%
Stature	F	1618	1690 ± 30	1677 ± 60	1645 ± 69	1687 ± 18
[mm]	M	1753	1690 ± 30	1745 ± 44	1785 ± 92	1693 ± 18
	F/M	92%	100%	96%	92%	100%
Seated height	F	844		879 ± 10	859 ± 24	851 ± 19
[mm]	M	901		886 ± 9	900 ± 41	875 ± 18
	F/M	94%		99%	95%	97%
Mass	F	62	66 ± 9	64 ± 6	61 ± 6	60 ± 12
[kg]	M	77	74 ± 9	76 ± 8	79 ± 14	75 ± 7
	F/M	81%	89%	84%	77%	80%

Vertebral body dimensions are smaller for females than males (Table 2), irrespecitve if measured on skeletal bone specimen or from medical images of patients and volunteers. This is expected as the female population is overall smaller than the male population. Frobin et al. (2002) found that the vertebral body height divided by depth ratio was smaller for females than males, indicating that the female cervical spine has less support area. When this ratio was calculated for other studies where both depth and height were reported, similar trends were seen (Table 2, last three columns). Vasavada et al. (2008) matched volunteers with regards to seated height and neck length, and found the female dimensions to be 86-98% of the male dimensions. Gender differences were most prominent for the C3-C5 vertebral bodies. DeRosia (2008) and Stemper et al. (2008, 2009) pusblished results with size matched volunteers in three groups based on either seated height, head circumference, or stature. They found that the female dimensions were 86-95% of the male dimensions and significantly smaller with regards to width and depth for females among the head circumference matched volunteers. The results from the head circumference matched group compare well with a large survey without size matching by Parenteau et al. (2014), which is likely because the head circumference matching provides a large spread in stature and other dimensions. Hence, even when size-matched, females have smaller vertebral bodies than males, which is most prominent for the vertebral depth.

A narrow spinal canal could possibly increase the risk of nervous tissue lesions. Significant gender differences in spinal canal dimensions have been found in studies of skeletal specimens and patients (Table 3). The spinal canal was widest in the upper cervical spine and narrowest at the C4-C6 levels. The largest gender differences were found at the C4-C5 level, where the spinal diameter was 4-6% narrower for females (Tatarek 2005 and Evangelopoulos et al. 2012, Parenteau et al. 2014). On the other hand, differences of less than 2% were reported for all levels (Hukuda and Kojima 2002) and for C4 and above (Parenteau et al. 2014). None of these studies comprised size-matched subjects and the gender differences may stem from a difference in overall size rather than gender.

Table 2. Vertebral Body Dimensions

F = females, M = males, No. = number of, skel. = skeletal specimen, vol = volunteers.

	Height*			Depth**									Width***					Height by Depth		
	average ± SD [mm]			average ± SD [mm]									average ± SD [mm]					[mm/mm]		
References	Boyle et al. 1996	Hukuda and Kojima 2002	Vasavada et al. 2008	Milne 1991	Kasai et al. 1996	Ebraheim et al. 1998	Hukuda and Kojima 2002	Vasavada et al. 2008	DeRosia 2008	Stemper et al. 2009	Stemper et al. 2009	Parenteau et al. 2014	Milne 1991	Vasavada et al. 2008	DeRosia 2008	Stemper et al. 2009	Stemper et al. 2009	Frobin et al. 2002	Hukuda and Kojima 2002	Vasavada et al. 2008
Size Matched	no	no	neck length	no	no	no	no	neck length	stature	sitting height	head circumf.	no	no	neck length	stature	sitting height	head circumf.	no	no	neck length
Subjects	skel.	skel.	vol.	skel.	skel.	skel.	skel.	vol.	vol.	vol.	vol.	patients	skel.	vol.	vol.	vol.	vol.	skel.	skel.	vol.
No. F	25	114	14	21	10	11	114	14	11	11	18	176	21	14	11	11	18	100	114	14
No. M	26	105	14	40		16	105	14	10	17	9	251	40	14	10	17	9	35	105	14
C1 F								45,1 ± 1,9												
M								47,6 ± 4,5												
F/M								95%												
C2 F						13,7 ± 1,3		13,1 ± 1,0	15,7 ± 1,1	15,9 ± 1,2	15,4 ± 0,9	14,4 ± 1,4			19,6 ± 2,3	20,4 ± 2,4	20,2 ± 1,7			
M						14,6 ± 1,2		14,2 ± 1,0	16,8 ± 0,9	17,5 ± 0,9	17,0 ± 0,8	16,7 ± 1,6			21,8 ± 1,0	22,3 ± 1,3	21,9 ± 1,0			
F/M						94%		92%	93%	91%	91%	86%			90%	91%	92%			
C3 F		13,0 ± 1,2	12,0 ± 0,9	13,8 ± 0,2		13,6 ± 1,4	16,0 ± 1,2	12,8 ± 0,8	15,4 ± 1,0	15,7 ± 0,8	15,1 ± 0,9	15,0 ± 1,6	18,2 ± 0,3	14,6 ± 2,6	20,4 ± 1,8	20,9 ± 1,9	20,7 ± 1,7	0,94	0,81	0,94
M		14,6 ± 1,5	13,5 ± 0,9	15,3 ± 0,2		14,6 ± 1,2	18,1 ± 1,3	14,2 ± 1,6	16,6 ± 1,0	16,8 ± 1,1	16,9 ± 1,2	17,2 ± 1,9	19,1 ± 0,3	16,9 ± 1,6	22,0 ± 1,3	22,9 ± 1,4	22,2 ± 1,9	0,91	0,81	0,95
F/M		89%	89%	90%		93%	88%	90%	93%	93%	89%	87%	95%	86%	93%	91%	93%	103%	101%	99%
C4 F		12,7 ± 1,2	11,6 ± 0,9	14,4 ± 0,3		13,5 ± 1,4	15,7 ± 1,3	13,0 ± 1,0	15,3 ± 0,9	15,7 ± 0,8	15,2 ± 1,0	15,3 ± 2,0	20,9 ± 0,3	16,9 ± 1,2	20,6 ± 2,0	21,9 ± 1,7	21,3 ± 1,5	0,88	0,81	0,89
M		14,2 ± 1,4	12,6 ± 0,8	16,0 ± 0,3		14,8 ± 1,6	17,6 ± 1,4	14,8 ± 1,2	16,9 ± 1,0	16,8 ± 1,1	17,1 ± 1,6	17,5 ± 2,1	19,3 ± 0,3	17,8 ± 2,1	22,2 ± 1,6	23,2 ± 1,3	22,7 ± 1,3	0,84	0,81	0,85
F/M		89%	92%	90%		91%	89%	88%	91%	93%	89%	87%	108%	95%	93%	94%	94%	105%	100%	105%
C5 F		12,7 ± 1,2	11,3 ± 0,7	14,0 ± 0,2	17,1 ± 0,6	13,4 ± 1,3	15,3 ± 1,3	13,0 ± 0,9	15,5 ± 0,9	15,8 ± 0,8	15,6 ± 1,2	16,6 ± 2,5	20,4 ± 0,3	18,3 ± 1,5	21,1 ± 1,5	22,1 ± 1,3	21,6 ± 1,8	0,85	0,83	0,87
M		13,9 ± 1,4	12,2 ± 0,6	16,2 ± 0,3	18,2 ± 0,5	14,6 ± 1,4	17,4 ± 1,4	15,2 ± 1,1	17,7 ± 1,2	16,9 ± 0,8	17,6 ± 1,2	18,7 ± 2,6	21,6 ± 0,2	19,0 ± 2,0	23,5 ± 1,7	23,8 ± 2,2	23,2 ± 1,5	0,81	0,80	0,80
F/M		91%	93%	86%	94%	92%	88%	86%	88%	93%	89%	89%	94%	96%	90%	93%	93%	104%	104%	108%
C6 F	11,1 ± 1,1	12,8 ± 1,3	11,6 ± 0,8	14,9 ± 0,5		13,3 ± 1,0	15,9 ± 1,3	14,0 ± 1,0	15,9 ± 0,9	16,3 ± 1,0	16,2 ± 1,0	16,9 ± 2,2	22,1 ± 0,4	19,8 ± 1,6	24,5 ± 1,8	24,7 ± 1,3	24,1 ± 1,8	0,83	0,81	0,83
M	11,8 ± 0,9	13,7 ± 1,2	12,0 ± 0,7	16,6 ± 0,3		14,8 ± 1,3	18,1 ± 1,4	15,6 ± 1,0	18,4 ± 2,0	18,0 ± 1,3	18,8 ± 2,0	19,0 ± 2,4	23,7 ± 0,3	20,9 ± 2,3	27,0 ± 1,6	26,0 ± 2,0	27,0 ± 1,2	0,85	0,76	0,77
F/M	94%	93%	97%	90%		92%	88%	90%	86%	91%	86%	89%	93%	95%	91%	95%	89%	99%	106%	108%
C7 F	12,9 ± 1,2	14,1 ± 1,2	12,8 ± 0,6	14,7 ± 0,3		12,9 ± 0,7	16,6 ± 1,3	13,7 ± 1,2				15,6 ± 1,7	23,4 ± 0,3	22,6 ± 2,4				0,91	0,85	0,93
M	13,3 ± 1,3	15,6 ± 1,5	13,0 ± 1,0	16,5 ± 0,3		14,3 ± 1,0	18,7 ± 1,2	15,5 ± 1,3				17,7 ± 1,9	25,2 ± 0,2	24,6 ± 2,7				0,92	0,83	0,84
F/M	97%	90%	98%	89%		90%	89%	88%				88%	93%	92%				98%	102%	111%
T1 F	14,3 ± 1,3			14,4 ± 0,3									23,4 ± 0,3							
M	15,3 ± 1,1			16,2 ± 0,2									25,2 ± 0,3							
F/M	94%			88%									93%							

* Height was measured at the middle of the vertebral body or the average of anterior and posterior height. ** Depth was measured at the middle of the vertbral body or the average of the superior and inferior surfaces, except for Milne 1991 (superior surface), Parenteau et al. 2014 (inferior surface), Stemper et al. 2009 and DeRosia 2008 (maximum measurement). *** Width was measured at the superior surface by Milne 1991, average of superior and inferior surface in Vasavada et al. 2008, and the maximum measurement in Stemper et al. 2009 and DeRosia 2008.

The facet joints are important for the biomechanical response. Gender differences in facet joint angles have been hypothesized as one explanation to the increased WAD risk for females. Measured relative to the vertebral body, very small and contradictory gender differences have been reported by Milne (1991), Boyle et al. (1996), Kasai et al. (1996), and Parenteau et al. (2013) (Table 3). Any difference in facet joint angle must thus stem from a difference in posture and/or spinal curvature. The variations in cervical spine alignment have been investigated extensively. For a majority of the population lordosis of the cervical spine is normal. The proportions of non-lordosis (straight or kyphosis) were 36% (Matsumoto et al. 1998) and 38% (Takeshima 2002) for an asymptomatic population measured in a seated position. Matsumoto et al. (1998) showed that females more frequently presented non-lordosis than males and that gender was an independent factor significantly associated with non-lordosis. Likewise, Helliwel et al .(1994) reported that the cervical spine was more likely to be straight in females than in males. Haedacker et al. (1997) observed that statistically cervical lordosis in males was more pronounced than in females. Additionally, Lee et al. (2014) focused on the cervicothoracic junction. Females had a thinner thoracic cage than males and males had a more forward-inclined thoracic inlet (angle between lines drawn from the top of the manubrium to the centroid of the cranial T1 end plate and the horizontal plane) than females. They concluded that a small anteroposterior diameter in the upper most thoracic cage was associated with cervical hypolordosis, whereas an inclined thoracic inlet was associated with pronounced cervical lordosis. Klinich et al. (2004) investigated the relationship between stature and type of cervical spine alignment. Tall females were more likely to have a straight cervical spine than short females. On the other hand, there was no significant relationship between stature and curvature for males. Hence, accounting for the spinal curvature, less lordosis will give a more horizontal orientation of the facet joint surfaces. An average female occupant HBM should include a seated posture with less lordosis than the average male models, based on these data. Also, the height of the female HBM should be considered when the seated lordosis is defined.

Dynamic response of occupants in rear impacts have been analysed with volunteer and post mortem human subject (PMHS) tests in order to understand the biomechanics of whiplash motion. In volunteer tests (Siegmund et al. 1997), females had greater head and neck kinematics compared to males in the overall motion analysis. In addition, cervical vertebral kinematics were also greater for females compared to males, when analysing sequential cineradiography images of cervical spines (Ono et al. 2006, Sato et al. 2014). Likewise, in rear impact sled tests with PMHS head-neck complexes, cervical vertebral kinematics were significantly greater for the female specimens than for the male specimens (Stemper et al. 2003, 2004). Initial alignment of cervical vertebrae has been considered as one of possible causes of the experimental gender difference seen. The alignment influences load transmission between the head and trunk of the body through the cervical spine, and can affect vertebral kinematics. Previous experimental studies (Maiman et al. 1983, Maiman et al. 2002, Yoganandan et al. 1986, Yoganandan et al. 1999, Liu and Dai 1989, Pintar et al. 1995) showed that changes in the initial alignment of the cervical spine had an influence on the severity of injury. Ono et al. (1997) conducted rear impact sled tests with a volunteer and quantified the effect of the initial alignment of the cervical spine on vertebral kinematics. They reported that rotational angles of cervical vertebrae were significantly greater in kyphotic alignment than in lordosis. Hence, as females are more likely to present with non-lordosis this implies that females inherently will suffer greater vertbral rotations. Likewise, Stemper et al. (2005) and Frechede et al. (2006) investigated elongation of the facet joint capsular ligaments (CL) in lordotic, straight and kyphotic cervical alignment with a FE head/neck model and reported that facet joint ligament elongations increased with kyphotic cervical alignment and concluded that kyphosispresented a higher risk for WAD than lordosis or straight spinal aligment.

Alongside anthropometry and spinal alignment, gender differences in material properties can influence the risk of lesions. Nightingale et al. (2002, 2007) tested upper and lower cervial spine specimen (4 female, 7 male) in flexion and extension bending load to failure. They found the mean upper cervical flexion and extension strength to be greater for males than females, however only significantly for flexion. The mean angle at failure were not significantly different for the genders. The female upper cervical spine was significantly less stiff than the male. Hence, the male range of motion was significantly greater than the female. Similar differences were seen also for the lower cervical spine. This, could be because of gender differences in cross sectional area of the ligaments or in the material properties. In sports, female athletes sustain more anterior cruciate ligament lesions (Elliot 2010), and some studies indicate that the influence of sex hormones on the material properties of the ligament may provide part of the explanation to increased risk (Wild 2012). Osakabe et al. (2001) studied the elastin and collagen content of lumbar ligaments with aging and found gender difference. The elastin content decreased with age for males but not for females. They suggested that sex steroid hormones may regulate the metabolism and thus effect how the elastin and collagen contents change with age. Stemper et al. (2010) tested thoracic specimen and found gender to influence the

elastic modulus in tension and compression, with female specimen having significantly greater moduli. Contrary, Bass et al. (2007) did not find any gender dependence for the failure true stress and failure true strain in cervical specimen (6 male, 5 female) tested for anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), and ligamentum flavum (LF). Mattucci et al. 2012 tested the ALL, PLL, LF, CL, and interspinous ligament (ISL) from 8 male and 8 female cadavers in quasi-static (0.5 s-1) and dynamic (up to 250 s-1) load rates. Stiffness and strength were consistently lower for female specimen, although significant gender differences were only found for the Young's modulus of the ALL and the failure force of the CL (Mattucci 2012). Hence, it seems likely that the gender difference due to anthropometry have a larger influence than gender difference in the material properties of the ligaments.

Table 3. Spinal Canal Diameter and Facet Joint Angles

None of these studies included size matched subjects. F = Females, M = Males, No. = Number of, skel. = skeletal specimen.
* Spinal canal anterior-posterior diameter in the sagittal plane. For Parenteau et al. 2014 the diameter was recalculated by doubling the reported radius. Tatarek 2005 values are the average of the two reported groups. ** The angle of the superior facet to the plane of the end plate, except for Parenteau et al. 2013 (inferior facet). Data from Kasai et al. 1996 was recalculated from the angle to the posterior vertebral line by adding 90 degrees.

		Spinal Canal Diameter *				**Facet Joint Angles ****			
		average ± SD [mm]				average ± SD [degrees]			
References		Hukuda and Kojima 2002	Tatarek 2005	Evangelopoulos et al. 2012	Parenteau et al. 2014	Milne 1991	Boyle et al. 1996	Kasai et al. 1996	Parenteau et al. 2013
Subjects		skel.	skel.	patients	patients	skel.	skel.	patients	patients
No. F		114	321	100	176	21	25	10	173
No. M		105			251	40	26		250
C1	F			16,3 ± 1,6					
	M			16,6 ± 1,9					
	F/M			98%					
C2	F		15,9 ± 1,4	13,3 ± 1,3	13,7 ± 1,6				126,5 ± 14,0
	M		16,6 ± 1,4	13,6 ± 1,6	13,7 ± 1,9				122,1 ± 12,3
	F/M		95%	97%	100%				104%
C3	F	16,8 ± 1,7	13,9 ± 1,4	12,9 ± 1,3	12,9 ± 1,5	126,2 ± 1,2		128,4 ± 3,8	127,0 ± 11,4
	M	16,6 ± 1,5	14,7 ± 1,3	13,3 ± 1,7	13,0 ± 1,7	122,8 ± 1,2		129,5 ± 4,2	124,1 ± 9,1
	F/M	101%	94%	97%	99%	103%		97%	102%
C4	F	15,8 ± 1,7	13,4 ± 1,4	12,5 ± 1,5	12,7 ± 1,5	124,8 ± 1,23		129,7 ± 2,8	128,4 ± 11,5
	M	15,8 ± 1,4	14,3 ± 1,3	13,1 ± 1,0	13,0 ± 1,9	124,5 ± 0,81		131,2 ± 4,1	129,0 ± 8,8
	F/M	100%	94%	96%	98%	100%		96%	100%
C5	F	15,7 ± 1,6	13,4 ± 1,3	12,7 ± 1,7	12,3 ± 1,7	125,7 ± 1,5		131,7 ± 3,0	128,0 ± 11,3
	M	15,9 ± 1,3	14,3 ± 1,3	13,4 ± 1,2	13,0 ± 1,6	127,6 ± 0,8		132,3 ± 2,5	129,3 ± 10,2
	F/M	99%	94%	94%	95%	99%		99%	99%
C6	F	15,6 ± 1,6	13,5 ± 1,1	12,5 ± 1,8	12,7 ± 1,5	124,8 ± 1,42	124,4 ± 5,7	125,3 ± 2,7	120,9 ± 11,9
	M	15,9 ± 1,3	14,3 ± 1,3	13,3 ± 1,9	13,3 ± 1,7	126 ± 0,89	125,7 ± 5,5	126,1 ± 4,5	124,5 ± 9,3
	F/M	98%	95%	94%	96%	99%	99%	98%	97%
C7	F	15,5 ± 1,6	13,5 ± 1,1		13,6 ± 1,5	116,0 ± 1,1	115,8 ± 5,7	115,2 ± 2,8	124,2 ± 14,4
	M	15,8 ± 1,4	14,4 ± 1,2		14,2 ± 1,4	115,6 ± 0,9	118,7 ± 5,1	117,4 ± 4,0	125,4 ± 11,4
	F/M	98%	94%		96%	100%	98%	92%	99%
T1	F					111,1 ± 1,13	108,5 ± 6,1		
	M					112,2 ± 1,05	112,2 ± 6,6		
	F/M					99%	97%		

FINITE ELEMENT HUMAN BODY MODELS

A number of whole body FE occupant models have been released, primarily of the 50th percentile male anthropometry as defined by Schneider et al. 1983. A first generation of models developed in the late 1990s and early 2000s are the THUMS v3 (Iwamoto et al. 2002), HUMOS (Robin 2001), and the JAMA model (Sugimoto and Yamazaki 2005). In the THUMS v3, 7 400 elements of the total 143 000 are located in the cervical region. However, quite substantial simplifications such as the the facet joints being represented by single hexahedral elements in shear, rather than a sliding contact, makes soft tissue injury prediction difficult. A second generation of whole body HBMs have more recently been released, with an increased number of elements by an order of

magnitude; the THUMS v4 (Watanabe et al. 2011) includes 1.7 million elements and the GHBMC model (Vavalle et al. 2013) 2.2 million elements to represent a 50th percentile male occupant. To date, the most advanced cervical model in a whole body HBM appears to be the GHBMC model, that includes sliding contact facet joints with articular cartilage, failure criteria for the interverterbral discs to represent disc avulsion, failure criteria for 1D cervical ligaments and bone, and 3D musculature with superimposed active 1D elements. For the GHBMC, 12% of the elements are located in the cervical region, while in the THUMS v4 only 1.6 % is used. For some of the models mentioned above, 95th percentile male and 5th percentile female versions have been developed and released. However, to the best of our knowledge, no 50th percentile female whole body HBM has been released to date.

Many models of the isolated cervical spine have been developed and used to study impact biomechanics problems. The first models employed multibody dynamics modelling techniques (de Jager 1996, Deng and Goldsmith 1987, van der Horst 2002), and were used to show the importance of musculature in capturing volunteer responses in multi-directional impacts (de Jager 1996, van der Horst 2002). The multibody approach is numerically more efficient than FE simulations, an important property for early models. When applied to whiplash motion, one major drawback of the multi body models is that the spinal joints were modelled with kinematic joints instead of facet surfaces restrained by soft tissues. One of the first FE models to inlude the complex joints of the upper cervical spine with sliding surfaces between the occiput and second cervical vertebra was Brolin and Halldin (2004), illustrating the benefit of this modelling approach when studying how material properites of ligaments influence kinematics. More recent studies have successfully been used to study the effect of cervical muscle activity, using optimisation (Chancey et al. 2003, Brolin et al. 2005, Dibb et al. 2013) or neuromuscular control schemes (de Bruijn 2014) for the cervical musculature. In order to capture the response of live human subjects through numerical simulation, detailed representation of the cervical musculature and at least a basic muscle activation scheme are necessary (Brolin et al. 2008, Hedenstierna and Halldin 2008). For rear-end impacts the importance of cervical muscle activation has also been confirmed through numerous experimental studies, e.g. by Siegmund et al. 2003.

The overwhelming majority of the developed impact biomechanics FE cervical spine models, summarised in Table 4, are based on the average male anthropometry. In general no stringent definition of the average male subject used for model development appear to have been used, such as defined for the GHBMC model (Gayzik et al. 2011), but rather a male volunteer or PMHS of arbitrary average size has provided the template geometry for the models. Interesting exceptions are the Duke University models, which have been scaled to represent the cervical spine of 6 and 10 year old children (Dibb et al. 2013), and the Nottingham Trent University model which was scaled to an average female size (Mordaka 2004). Although a limited number of publications were made with the average female model, it was concluded that scaling alone is not sufficient to capture gender differences in head-neck kinematics during rear-end impact simulations (Mordaka 2004). Furthermore, an expected trend of increasing number of elements with later years can be found, with the exception of the parametric model of Laville et al.(2009). By parameterizing and simulating 16 different individual cervical spine geometries, almost all variation present in experimental moment-rotation corridors for the C5–C6 spinal unit was captured, although material properties were kept constant (Laville et al. 2009). As shown by the review of female anthropometry in this paper, significant differences between female and male cervical vertebral geometry exist; hence, the findings of Laville et al. 2009 support the need for HBMs based on female anthropometry.

The strength of the FE method for impact biomechanics modelling is that it allows for localised injury prediction, rather than using global kinematic and force criteria, as used for assessment with anthropomorphic test devices. In about half of the publications included in Table 4, no assessment of soft tissue injury has been considered. For the rest, the most commonly assessed parameter is ligament strain, which can either be relative between simulations (Brolin et al. 2008, Kitagawa et al. 2008), or compared with experimentally recorded failure levels (Cronin et al. 2012, DeWit and Cronin 2012, Fice and Cronin 2012, Fice et al. 2011, Panzer et al. 2011). Peak intervertebral annulus fibrosus strain has been considered (Panzer et al. 2011), as well as shear deformation of the disc (Fice et al. 2011), the change in intervertebral disc pressure and cross-sectional force between simulations (Brolin et al. 2008), and disc avulsion by tied contact failure (Cronin et al. 2012). For the hard tissues, bone fracture simulation by element elimination is quite common (Cronin et al. 2012, DeWit and Cronin 2012, Halldin et al. 2000, Panzer et al. 2011, Zhang et al. 2005), although some questions about post failure response accuracy have been raised (DeWit and Cronin 2012). Furthermore, global injury criteria that have been suggested, or used as complements to local criteria, were the neck injury criterion (Kitagawa et al. 2008) and cervical spine tensile or compressive forces (Camacho et al. 1997, Dibb et al. 2013).

Table 4. Summary of FE Cervical Spine Models for Impact Biomechanics

Anthrop. = Anthropometry, No. Elem. = Number of elements, M50 = average male, F50 = average female, PMHS = Post Mortem Human Subject. CT = Computed Tomography. MRI = Magnetic Resonance Imaging. 1,2,3D = One, two, or three dimensional elements. *Only elements in the cervical spine are reported here. The number of elements may vary with different versions of the models.

References	Anthrop.	Skeletal geometry	No. Elem.*	Solver	Facet Joints	Ligaments	Muscles
Camacho et al. 1997 Chancey et al. 2003 Dibb et al. 2013	M50 6-year-old 10-year-old	Visible Human Database. Child models scaled based on anatomical literature.	1137	LS-DYNA	Non-linear kinematic joints govern intervertebral joint motion, soft tissues not modelled individually.		Active 1D
Yang et al. 1998	M50	Volunteer MRI data.	14700	PAM-CRASH	Sliding contact	Non-linear 1D and 2D	Not included
Halldin et al. 2000 Brolin and Halldin 2004 Brolin et al. 2005 Brolin et al. 2008 Hedenstierna et al. 2008	M50	Volunteer CT data.	8400	LS-DYNA	Sliding contact	Non-linear 1D	Active 1D and 3D
Jost and Nurick 2000	M50	Anatomical literature.	3000	ABAQUS	Not included	2D	Passive 2D
Gentle et al. 2001 Mordaka 2004	M50 F50	Visible Human Database. F50 scaled from M50 based on anatomical literature.	Not reported	LS-DYNA	Not reported	Non-linear 1D and 3D	Passive 1D
Meyer et al. 2004	M50	Volunteer CT data.	73185	RADIOSS	Sliding contact	1D	Passive 3D
Zhang et al. 2005 Zhang et al. 2006	M50	Volunteer external measurements.	22094	LS-DYNA	Sliding contact	Non-linear 1D	Not included
Eijima et al. 2005	M50	JAMA model [x] and volunteer MRI scan	14300	LS-DYNA	Solid elements	2D	Passive 3D
Panzer 2006 Panzer and Cronin 2009 Fice et al. 2011 Panzer et al. 2011 Cronin et al. 2012 DeWit and Cronin 2012 Fice and Cronin 2012	M50	Commercial geometry package.	108300	LS-DYNA	Sliding contact with synovial fluid	Non-linear 1D with failure	Active 1D
Wheeldon et al. 2008	M50	PMHS CT data and cryomicrotome data.	Not reported	ABAQUS	Sliding contact with synovial fluid	Non-linear 1D	Not included
Kitagawa et al. 2008	M50	From THUMS v.4 (Watanabe et al. 2011)	80000	LS-DYNA	Sliding contact with synovial fluid	2D	Passive 1D
Laville et al. 2009	Subject specific	Volunteer CT	5350	RADIOSS	Sliding contact	Non-linear 1D	Not included
Cervical spine models in whole human body models							
Iwamoto et al. 2002 THUMS v3	M50	Commercial geometry package.		LS-DYNA,PAM-CRASH	Solid element in shear	Linear 2D	Passive 1D
Watanabe et al. 2011 THUMS v4	M50	CT Scan data set.		LS-DYNA, PAM-CRASH	Solid element in shear	Linear 2D	Passive 1D
Robin et al. 2001 HUMOS/HUMOS2	M50	Photos of PMHS slices.		MADYMO, RADIOSS, PAM-CRASH	Sliding contact	Non-linear 1D	Passive 1D
Vavalle et al. 2013 GHBMC	M50	Volunteer CT and MRI data sets.		LS-DYNA	Sliding contact	Non-linear 1D with failure	Active 1D
Sugimoto and Yamazaki 2005, JAMA	M50	Based on other FE models.		LS-DYNA, PAM-CRASH	Unknown	Unknown	Unknown

In relation to proposed injury mechanisms for WAD, summarised in the Injury Mechanisms section, synovial fold pinching, tension of dorsal root ganglia or vertebral arteries, and musculoskeletal injury appear not to have been covered in detail in studies with FE impact biomechanics models of the cervical spine. It is likely that further model development is nescessary to assess these mechanisms. Another mechanism not investigated by the models summarised in Table 4 is cervical canal pressure transients that may load and deform nerve tissue such as the spinal ganlia, which require coupled fluid dynamic and solid mechanics simulations in the HBM or in a submodel that use the output from the HBM as input in order to model local phenomena.

SUMMARY AND CONCLUSIONS

Real-world data has identified the female occupant to be at more risk for WAD in traffic accidents and many studies illustrate important differences in the kinematic response of male and female subjects, volunteers and PMHSs. The review of FE models of the cervical spine presented 17 models based on male geometry but only one model that was scaled to represent a female. It is obvious that current tools are not adequate for assessing seat performance or develop protection against WAD with the female population in focus. An average female HBM has the potential to improve WAD prevention and assessment of seat performance to reduce traffic injuries in the population most at risk of these injuries. When the EvaRID model was developed, the average female was defined on the anthropometric measures of the 50th percentile female from the UMTRI study (stature 161.8 cm, mass 62.3 kg, Schneider et al. 1983) as it was considered a reasonable representation of the worldwide population at risk (Carlsson 2014). Another advantage of using the data by Schneider et al. (1983) for the average female HBM is that it would be consistent with several of the average male models developed based on the 50th percentile male (stature 175.3 cm, 77.3 kg, , Schneider et al. 1983). This approach would provide stronger comparisons between average male and female responses for seat assessment and other safety applications.

The literature review leaves no doubt that female HBMs should be developed directly based on female anthropometry and not scaled from male data because females have more slender necks and less muscle mass than males. Gender significantly influences the shape of the vertebral body, with the female vertebral body being less deep and not as wide. There was no or very little influence of gender on the height of the vertebral bodies and the diameter of the spinal canal. The facet joint angles compared to a reference in the vertebral body did not seem to depend on gender. Instead, gender significantly influenced the spinal alignment in the seated posture, where females on average present a straighter spine and males show more lordosis. There may be an effect of gender on the material properties of the ligaments in the cervical spine; however it is probably less important than gender differences in the cross sectional area of ligaments. In future studies on gender dependence, using size matched male and female subjects based on neck length or seated height rather than head circumference is recommended to distinguish the effect of gender from size.

To conclude, we recommend that average female HBMs are developed with focus on;

- The shape of the female vertebral body, especially the depth and width that provides less support area than for males,
- Defining a spinal curvature representative of seated female volunteers that on average present with less lordosis than males,
- The dimensions of the spinal ligaments, rather than the material properties, to capture the larger range of motion and less spinal stiffness of female subjects compared to male, and
- Validation of female volunteers and PMHS tests for range of motion, while failure prediction seem to be less gender sensitive.

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REFERENCES

Banic B, Petersen-Felix S, Andersen O, Radanov B, Villiger P, Arendt-Nielsen L, Curatolo M. (2004) Evidence for Spinal Cord Hypersensitivityin Chronic Pain after Whiplash Injury and in Fibromyalgia. Pain 107: 7–15.

Bass CR, Lucas SR, Salzar RS, Oyen ML, Planchak C, Shender BS, Paskoff G (2007) Failure Properties of Cervical Spinal Ligaments under Fast Strain Rate Deformations. Spine 32(1): E7–13.

Boyle J, Singer K, Milne N. (1996) Morphological survey of the cervicothoracic junctional region. Spine 21(5): 544-548

Brault JR, Siegmund GP, Wheeler JB. (2000) Cervical Muscle Response during Whiplash: Evidence of a Lengthening Muscle Contraction. Clin. Biomech. 15: 426–435.

Brolin K, Halldin P. (2004). Development of a Finite Element Model of the Upper Cervical Spine and a Parameter Study of Ligament Characteristics, Spine 29(4): 376-385.

Brolin K, Halldin P, Leijonhufvud I. (2005) The Effect of Muscle Activation on Neck Response, Traffic Inj Prev. 6(1): 67–76.

Brolin K, Hedenstierna S, Halldin P, Bass C, Alem N. (2008). The Importance of Muscle Tension on the Outcome of Impacts with a Major Vertical Component. International Journal of Crashworthiness 13(5): 487–498.

Camacho D, Nightingale R, Robinette J, Vanguri S, Coates D, Myers B. (1997). Experimental Flexibility Measurements for the Development of a Computational Head-Neck Model Validated for Near-Vertex Head Impact, In Proceedings of the 41st Stapp Car Crash Conference (Orlando, FL, USA). Paper no. 973345.

Carlsson A, Chang F, Lemmen P, Kullgren A, Schmitt KU, Linder A, Svensson M. (2014) Anthropometric Specifications, Development, and Evaluation of EvaRID - A 50th Percentile Female Rear Impact Finite Element Dummy Model. Traffic Inj Prev. 15(8): 855-865.

Carstensen T, Frostholm L, Oernboel E, Kongsted A, Kasch H, Jensen T, Fink P (2012) Are There Gender Differences in Coping with Neck Pain Following Acute Whiplash Trauma? A 12-Month Follow-Up Study, Eur J Pain. 16(1): 49–60.

Chancey V, Nightingale R, Van Ee C, Knaub K, Myers B. (2003) Improved Estimation of Human Neck Tensile Tolerance: Reducing the Range of Reported Tolerance Using Anthropmetrically Correct Muscles and Optimized Physilogic Initial Conditions, Stapp Car Crash J. 47: 135–153.

Chapline J, Ferguson S, Lillis R, Lund A, Williams A (2000) Neck Pain and Head Restraint Position Relative to the Driver's Head in Rear-End Collisions, Accid Anal Prev. 32(2): 287–297.

Cronin D, Fice J, DeWit J, Moulton J. (2012) Upper Cervical Spine Kinematic Response and Injury Prediction, In Proceedings of the IRCOBI Conference (Dublin, Ireland, Sept. 12–14), 225–234.

Curatolo M, Bogduk N, Ivancic P, McLean S, Siegmund G, Winkelstein B (2011) The role of tissue damage in whiplash associated disorders: Discussion paper 1. Spine. 36(25 Suppl): S309–315.

Curatolo M, Petersen-Felix S, Arendt-Nielsen L, Giana C, Zbinden AM, Radanov BP. (2001) Central Hypersensitivity in Chronic Pain after Whiplash Injury. Clin J Pain. 17(4): 306–315.

Davidsson, J., Lövsund, P., Ono, K., Svensson, M. Y., & Inami, S. (1999). A comparison between volunteer, BioRID P3 and Hybrid III performance in rear impacts. In Proceedings of the International Research Council on the Biomechanics of Injury conference (Vol. 27, pp. 165-178). International Research Council on Biomechanics of Injury.

de Bruijn E. (2014) Isometric and Dynamic Control of Neck Muscles – Reflexive Contributions and Muscle Synergies, Doctoral Thesis, Delft University of Technology, Delft, the Netherlands.

de Jager M. (1996) Mathematical Head-Neck Models for Acceleration Impacts, Doctoral Thesis, Eindhoven University of Technology, Eindhoven, the Netherlands.

Deng Y-C, Goldsmith W. (1987) Response of a Human Head/Neck/Upper-Torso Replica to Dynamic Loading – II. Analytical/Numerical Model, J Biomech. 20(5): 487–497.

DeRosia J (2008) Role of Gender and Size in Biomechanics of Rear Impact. Doctoral Dissertation, Marquette University, USA, UMI Number: 3357947.

DeWit J, Cronin D (2012) Cervical Spine Segment Finite Element Model for Traumatic Injury Prediction, J Mech Behav Biomed Mater. 10: 138–150.

Dibb A, Cox C, Nightingale R, Luck J, Cutcliffe H, Myers B, Arbogast K, Seacrist T, Bass C. (2013) Importance of Muscle Activation for Biofidelic Pediatric Neck Response in Computational Models", Traffic Inj Prev. 14 Suppl: S116–127.

Dolinis J (1997) Risk Factors for "Whiplash" in Drivers: a Cohort Study of Rear-End Traffic Crashes, Injury 28(3): 173–179.

Ebraheim N, Fow J, Xu R, Yeasting R. (1998) The Vertebral Body Depths of the Cervical Spine and Its Relation to Anterior Plate-Screw Fixation. Spine 23(21): 2299-2302.

Eichberger A, Geigl BC, Moser A, Fachbach B, Steffan H. (1996) Comparison of Different Car Seats Regarding Head-Neck Kinematics of Volunteers During Rear End Impact. Proceedings of the International Research Conference on the Biomechanics of Impact, IRCOBI, Dublin (Ireland).

Ejima S, Ono K, Kaneoka K, Fukushima M. (2005) Development and Validation of the Human Neck Muscle Model Under Impact Loading, In Proceedings of the IRCOBI Conference (Prague, Czech Republic, Sept. 21–23), 245–255.

Elliot D, Goldberg L, Kuehl K. (2010) Young Women's Anterior Cruciate Ligament Injuries An Expanded Model and Prevention Paradigm. Sports med 40(5): 367-376.

Evangelopoulos D, Kontovazenitis P, Kouris S, Zlatidou X, Benneker L, Vlamis J, Korres D, Efstathopoulos N. (2012) Computerized Tomographic Morphometric Analysis of the Cervical Spine. Open Orthop J. 6: 250-254.
Fice J, Cronin D. (2012) Investigation of Whiplash Injuries in the Upper Cervical Spine using a Detailed Neck Model, J Biomech. 45(6): 1098–1102.
Fice J, Cronin D, Panzer M. (2011) Cervical Spine Model to Predict Capsular Ligament Response in Rear Impact, Ann Biomed Eng. 39(8): 2152–2162.
Frechede B, Bertholon N, Saillant G, Lavaste F, Skalli W (2006) Finite Element Model of the Human Neck during Omni-directional Impacts. Part II: Relation between Cervical Curvature and Risk of Injury. Comput Methods Biomech Biomed Engin. 9(6): 379-386.
Frobin W, Leivseth G, Biggemann M, Brinckmann P. (2002) Vertebral height, disc height, posteroanterior displacement and dens–atlas gap in the cervical spine: precision measurement protocol and normal data. Clin Biomech. 17: 423-431.
Gayzik F, Moreno D, Geer C, Wuertzer S, Martin R, Stitzel J. (2011). Development of a Full Body CAD Dataset for Computational Modeling: A Multi-Modality Approach, Ann Biomed Eng. 39(10): 2568–2583.
Gentle, C.R., Golinski, W.Z., and Heitplatz, F. (2001) Computational Studies of 'Whiplash' Injuries, Proc Inst Mech Eng H. 215(2): 181–189.
Haedacker JW, Shuford RF, Capicoto PN, Pryor PW (1997) Radiographic Standing Cervical Segmental Alignment in Adult Volunteers without Neck Symptoms. Spine. 22(13): 1472-1480.
Halldin P, Brolin K, Kleiven S, von Holst H, Jakobsson L, Palmertz C. (2000) Investigation of Conditions that Affect Neck Compression-Flexion Injuries Using Numerical Techniques, In Proceedings of the 44th Stapp Car Crash Conference (Atlanta, GA, USA, Nov. 6–8). Paper no. 2001-01-SC10.
Harty J, Quinlan J, Kennedy J, Walsh M, O'Byrne J. (2004) Anthropometrical analysis of cervical spine injuries. Injury 35(3): 249-252.
Hedenstierna S, Halldin P. (2008) How Does a Three-Dimensional Continuum Muscle Model Affect the Kinematics and Muscle Strains of a Finite Element Neck Model Compared to a Discrete Muscle Model in Rear-End, Frontal, and Lateral Impacts, Spine. 33(8): E236–245.
Helliwel PS, Evans PF, Wright V (1994) The Straight Cervical Spine: Does it Indicate Muscle Spasm? J Bone Joint Surg Br. 76(1): 103-106.
Hukuda S, Kojima Y. (2002) Sex Discrepancy in the Canal/Body Ratio of the Cervical Spine Implicating the Prevalence of Cervical Myelopathy in Men. Spine 27(3): 250-253.
Iwamoto M, Kisanuki Y, Watanabe I, Furusu K, Miki K. (2002) Development of a Finite Element Model of the Total Human Model for Safety (THUMS) and Application to Injury Reconstruction, In Proceedings of the IRCOBI Conference (Munich, Germany, Sep. 18–20)
Jakobsson L, Norin H, Svensson M. (2004) Parameters Influencing AIS 1 Neck Injury Outcome in Frontal Impacts. Traffic Inj Prev. 5(2): 156–163.
Jost R, Nurick G. (2000) Development of a Finite Element Model of the Human Neck Subjected to High G-Level Lateral Deceleration, International Journal of Crashworthiness, 5(3): 259–269.
Kaneoka K, Ono K, Inami S, Hayashi K (1999) Motion analysis of cervical vertebrae during whiplash loading. Spine. 24(8): 763-769.
Kasai T, Ikata T, Katoh S, Miyake R, Tsubo M. (1996) Growth of the Cervical Spine With Special Reference to Its Lordosis and Mobility. Spine 21(18): 2067-2073.
Kasch H, Stengaard-Pedersen K, Arendt-Nielsen L, Staehelin Jensen T. (2001) Pain Thresholds and Tenderness in Neck and Head Following Acute Whiplash Injury: A Prospective Study. Cephalalgia 21(3): 189–197.
Kihlberg JK (1969) Flexion-Torsion Neck Injury in Rear Impacts. Proceedings of the 13th Association for the Advancement of Automotive Medicine (AAAM) Conference, pp. 1–16.
Kitagawa Y, Yasuki T, Hasegawa J. (2008) Research Study on Neck Injury Lessening with Active Head Restraint Using Human Body FE Model", Traffic Inj Prev. 9(6): 574–582.
Klinich KD, Ebert SM, Van Ee CA, Flannagan CAC, Prasad M, Reed MP, Schneider LW (2004) Cervical Spine Geometry in the Automotive Seated Posture: Variations with Age, Stature , and Gender. Stapp Car Crash J. 48: 301-330.
Krafft M. (1998). Non-fatal injuries to car occupants, Injury assessment and analysis of impacts causing short- and long-term consequences with special reference to neck injuries. Doctoral thesis, Karolinska Institutet, Stockholm, Sweden.
Krafft M, Kullgren A, Lie A, Tingvall C. (2003) The Risk of Whiplash Injury in the Rear Seat Compared to the Front Seat in Rear Impacts, Traffic Inj Prev. 4(2): 136–140.
Krakenes J, Kaale BR. (2006) Magnetic Resonance Imaging Assessment of Craniovertebral Ligaments and Membranes after Whiplash Trauma. Spine. 31(24): 2820–2826.
Kullgren A, Krafft M, Tingvall C, Lie A. (2003) Combining crash recorder and paired comparison technique: Injury risk functions in frontal and rear impacts with special reference to neck injuries. Proc. 18th ESV Conf, Paper No. 404, Nagoya (Japan).
Kullgren A, Krafft M, Lie A, Tingvall C (2007) The Effect of Whiplash Protection Systems in Real-Life Crashes and Their Correlation to Consumer Crash Test Programmes, Proc. 20th ESV Conf. (07-0468), Lyon (France).

Kullgren A, Stigson H, Krafft M (2013) Development of Whiplash Associated Disorders for Male and Female Car Occupants in Cars Launched Since the 80s in Different Impact Directions. Proceedings of the International Research Conference on the Biomechanics of Impact, IRCOBI, pp 51-62.
Laville A, Laporte S, Skalli W. (2009) Parametric and Subject-Specific Finite Element Modelling of the Lower Cervical Spine. Influence of Geometrical Parameters on the Motion Patterns. J Biomech. 42(10): 1409–1415.
Lee JH, Park YK, Kim JH (2014) Chronic Neck Pain in Young Adults: Perspectives on Anatomic Differences. Spine J. 14(11): 2628-2638.
Linder A, Schick S, Hell W, Svensson M, Carlsson A, Lemmen P, Schmitt K-U, Gutsche, A, Tomasch E. (2013) ADSEAT – Adaptive Seat to Reduce Neck Injuries for Female and Male Occupants, Accid Anal Prev. 60: 334-343.
Liu YK, Dai QG (1989) The Second Stiffest Axis of a Beam-Colum: Implications for Cervical Spine Trauma. J Biomech Eng. 111(2): 122-127.
Maag U, Desjardins D, Bourbeau R, Laberge-Nadeau C (1990) Seat Belts and Neck Injuries, Proceedings of the IRCOBI Conference, Bron (France), pp. 1–14.
Maiman D, Sances A Jr, Myklebust J, Larson S, Houterman C, Chilbert M, El-Ghatit A (1983) Compression Injury of the Cervical Spine. Neurosurgery. 13(3): 254-260.
Maiman D, Yoganandan N, Pintar F (2002) Preinjury Cervical Alignment Affecting Spinal Trauma. J Neurosurg. 97(1 Suppl): 57-62.
Matsumoto M, Fujimura Y, Suzuki N, Toyama Y, Shiga H (1998) Cervical Curvature in Acute Whiplash Injures: Prospective Comparative Study with Asymptomatic Subjects. Injury. 29(10): 775-778.
Mattucci S, Moulton J, Chandrashekar N, Cronin D. (2012) Strain rate dependent properties of younger human cervical spine ligaments. J Mech Behav Biomed Mater. 10: 216-226.
Meyer F, Bourdet N, Deck C, Willinger R, Raul J. (2004) Human Neck Finite Element Model Development and Validation against Original Experimental Data, Stapp Car Crash J. 48, Paper no. 2004–22–0008.
Milne N. (1991) The role of zygapophysial joint orientation and uncinated processes in controlling motion in the cervical spine. J. Anat. 178: 189-201.
Mordaka J. (2004) Finite Element Analysis of Whiplash Injury for Women, Doctoral Thesis, Nottingham Trent University, Nottingham, UK.
Morris AP, Thomas PD (1996) Neck Injuries in the UK Co-operative Crash Injury Study, Proc. 40th Stapp Car Crash Conf., SAE 962433), pp. 317–329.
Narragon EA (1965) Sex Comparisons in Automobile Crash Injury, Cornell Aeronautical Laboratory (CAL) Report No. VJ 1823-R15.
Nightingale R, Chancey C, Ottaviano D, Luck J, Tran L, Prange M, Myers B. (2007) Flexion and extension structural properties and strengths for male cervical spine segments. J Biomech 40(3): 535-542.
Nightingale R, Winkelstein B, Knaub K, Richardson W, Luck J, Myers B. (2002) Comparative strengths and structural properties of the upper and lower cervical spine in flexion and extension. J Biomech 35(6): 725-732.
Nygren, Å. (1983). Injuries to Car Occupants - Some Aspects of the Interior Safety of Cars. Acta Oto-Laryngologica. 95(s395): 1-135, doi: 10.3109/00016488309139642.
O'Neill B, Haddon W Jr, Kelley AB, Sorenson WW (1972) Automobile Head Restraints: Frequency of Neck Injury Claims in Relation to the Presence of Head Restraints. Am J Public Health. 62(3): 309–406.
Ono K, Ejima S, Suzuki S, Kaneoka K, Fukushima M, Ujihashi S. (2006) Prediction of Neck Injury Risk based on the Analysis of Localized Cervical Vertebral Motion of Human Volunteers during Low-speed Rear Impacts. Proceedings of IRCOBI Conference, Madrid, Spain.
Ono K, Kaneoka K, Wittek A, Kajzer J. (1997) Cervical Injury Mechanism Based on the Analysis of Human Cervical Vertebral Motion and Head-Neck-Torso Kinematics during Low-speed Rear Impacts. Proc.41st Stapp Car Crash Conference, pp. 339–356.
Örtengren T, Hansson HA, Lövsund P, Svensson MY, Suneson A, Säljö A. (1996) Membrane Leakage in Spinal Ganglion Nerve Cells Induced by Experimental Whiplash Extension Motion: A Study in Pigs. J Neurotrauma. 13(3): 171–180.
Osakabe T, Hayashi M, Hasegawa K, Okuaki T, Ritty T, Mecham R, Wachi H, Seyama Y. (2001) Age- and Gender-Related changes in ligament components. Ann Clin Biochem. 38(Pt 5): 527- 532.
Otremski I, Marsh JL, Wilde BR, McLardy Smith PD, Newman RJ (1989) Soft tissue cervical injuries in motor vehicle accidents. Injury. 20(6): 349–351.
Panjabi MM. (2006) A Hypothesis of Chronic Back Pain: Ligament Subfailure Injuries Lead to Muscle Control Dysfunction. Eur Spine J. 15(5): 668–676.
Panzer M. (2006) Numerical Modeling of the Human Cervical Spine in Frontal Impact. Master Thesis, University of Waterloo, Waterloo, ON, Canada.
Panzer M, Cronin D. (2009) C4–C5 Segment Finite Element Model Development, Validation, and Load-Sharing Investigation, J Biomech. 42(4): 480–490.
Panzer M, Fice J, Cronin D. (2011) Cervical Spine Response in Frontal Crash, Med Eng Phys. 33(9): 1147–1159.
Parenteau C, Wang N, Zhang P, Caird M, Wang S. (2014) Quantification of Pediatric and Adult Cervical Vertebra—Anatomical Characteristics by Age and Gender for Automotive Application. Traffic Inj Prev. 15(6): 572-582.

Parenteau C, Zuby D, Brolin K, Svensson M, Palmertz C, Wang S. (2013) Restrained Male and Female Occupants in Frontal Crashes: Are We Different? International IRCOBI Conference on the Biomechanics of Impact, September 11-13, Gothenburg, Sweden.

Pearson AM, Ivancic PC, Ito S, Panjabi MM. (2004) Facet Joint Kinematics and Injury Mechanisms during Simulated Whiplash. Spine. 29(4): 390–397.

Pintar F, Yoganandan N, Voo L, Cusick JF, Maiman DJ, Sances A Jr (1995) Dynamic Characteristics of the Human Cervical Spine. 39th Stapp Car Crash Conference Proceedings, pp. 195-202.

Reddy M, Reddy B, Schoggl A, Saringer W, Matula C. (2002) The Complexity of Trauma to the Cranio-Cervical Junction: Correlation of Clinical Presentation with Doppler Flow Velocities in the V3- Segment of the Vertebral Arteries. Acta Neurochir (Wien). 144(6): 575–580.

Richter M, Otte D, Pohlemann T, Krettek C, Blauth M (2000) Whiplash-Type Neck Distortion in Restrained Car Drivers: Frequency, Causes and Long-Term Results, Eur Spine J. 9(2): 109–117.

Robin S. (2001) HUMOS: Human Model for Safety – A Joint Effort Towards the Development of Refined Human-like Car Occupant Models. 17th ESV Conference (Amsterdam, the Netherlands, June 4–7). Paper no. 297.

Sato F, Nakajima T, Ono K, Svensson M, Brolin K, Kaneoka K (2014) Dynamic Cervical Vertebral Motion of Female and Male Volunteers and Analysis of its Interaction with Head/Neck/Torso Behavior during Low-speed Rear Impact. Proceedings of IRCOBI Conference, Berlin, Germany.

Schmitt K-U, Weber T, Svensson M, Davidsson J, Carlsson A, Björklund M, Jakobsson L, Tomasch E, Linder A (2012) Seat testing to investigate the female neck injury risk - preliminary results using a new female dummy prototype, Proceedings of the International Research Conference on the Biomechanics of Impact, IRCOBI, Dublin (Ireland).

Schneider LW, Robbins DH, Pflüg MA, Snyder RG. (1983) Development of Anthropometrically Based Design Specifications for an Advanced Adult Anthropomorphic Dummy Family, Final Report, UMTRI-83-53-1, University of Michigan Transportation Research Institute, Ann Arbor, Michigan, USA.

Scott D, Jull G, Sterling M. (2005) Widespread Sensory Hypersensitivity Is a Feature of Chronic Whiplash-Associated Disorder but Not Chronic Idiopathic Neck Pain. Clin J Pain. 21(2): 175–181.

Seric V, Blazic-Cop N, Demarin V. (2000) Haemodynamic Changes in Patients with Whiplash Injury Measured by Transcranial Doppler Sonography (TCD). Coll Antropol. 24(1): 197–204.

Siegmund GP, King DJ, Lawrence JM, Wheeler JB, Brault JR, Smith TA (1997) Head/neck kinematic response of human subjects in low-speed rear-end collisions, Proceedings of 41th Stapp Car Crash Conference, Lake Buena Vista, FL, USA, SAE paper 973341.

Siegmund G, Winkelstein B, Ivancic P, Svensson M, Vasavada A (2009) The anatomy and biomechanics of acute and chronic whiplash injury. Traffic Inj Prev. 10(2): 101-112.

Siegmund G, Sanderson D, Myers B, Inglis J. (2003) Rapid Neck Muscle Adaptation Alters the Head Kinematics of Aware and Unaware Subjects Undergoing Multiple Whiplash-Like Perturbations. J Biomech. 36(4): 473–482.

Stemper B, Board D, Yoganandan N, Wolfla C. (2010) Biomechanical properties of human thoracic spine disc segments. J Craniovertebr Junction Spine 1(1): 18-22.

Stemper B, DeRosia J, Yogananan N, Pintar F, Shender B, Paskoff G (2009) Gender Dependent Cervical Spine Anatomical Differences in Size-Matched Volunteers. 46th International ISA Biomedical Science Instrumentation Symposium, Milwaukee WI, 17-19 April, 149-154.

Stemper B, Yoganandan N, Pintar F. (2005) Effects of Abnormal Posture on Capsular Ligament Elongations in a Computational Model Subjected to Whiplash Loading. J Biomech. 38(6): 1313–1323.

Stemper B, Yoganandan N, Pintar FA. (2004) Gender- and Region- Dependent Local Facet Joint Kinematics in Rear Impact. Spine 29(16): 1764–1771.

Stemper B, Hallman J, Pintar F, Maiman D. (2012) Gender and Aging: Considerations for Orthopaedics. In Orthopaedic Biomechanics, Editor: Beth A. Winkelstein, Published: December 18, 2012 by CRC Press

Stemper B, Yoganandan N, Pintar FA, Maiman DJ, Meyer MA, DeRosia J, Shender BS, Paskoff G (2008) Anatomical Gender Differences in Cervical Vertebrae of Size-Matched Volunteers. Spine. 33(2): E44–E49.

Stemper B, Yoganandan N, Pintar F. (2003) Gender Dependent Cervical Spine Segmental Kinematics during Whiplash. J Biomech. 36(9): 1281–1289.

Storvik S, Stemper B, Yoganandan N, Pintar F. (2009) Population-Based Estimates of Whiplash Injury Using NASS CDS Data, Biomed Sci Instrum. 45: 244–249.

Sugimoto T, Yamazaki K. (2005). First Results from the JAMA Human Body Model Project. 19th ESV Conf. (Washington D.C., US), Paper no. 05–0291.

Svensson MY, Boström O, Davidsson J, Hansson H-A, Håland Y, Lövsund P, Suneson A, Säljö A. (2000) Neck Injuries in Car Collisions—A Review Covering a Possible Injury Mechanism and the Development of a New Rear-Impact Dummy. Accid Anal Prev. 32(2): 167–175.

Takeshima T, Omokawa S, Takaoka T, Araki M, Ueda Y, Takakura Y (2002) Sagittal Alignment of Cervical Flexion and Extension. Spine. 27(15): E348–355.

Tatarek N. (2005) Variation in the human cervical neural canal. Spine J. 5(6): 623-631.

Temming J, Zobel R (1998) Frequency and Risk of Cervical Spine Distortion Injuries in Passenger Car Accidents: Significance of Human Factors Data, Proceedings of the IRCOBI Conference, Göteborg, Sweden, pp. 219–233.

The Whiplash Commission (2005) *Final Report* http://www.whiplashkommissionen.se/pdf/WK_finalreport.pdf.

Thomas C, Faverjon G, Hartemann F, Tarriere C, Patel A, Got C (1982) Protection Against Rear-End Accidents, Proceedings of the IRCOBI Conference, Cologne, Germany, pp. 17–29.
Valkeinen H, Ylinen J, Malkia E, Alen M, Hakkinen K. (2002) Maximal force, force/time and activation /coactivation characteristics of the neck muscles in extension and flexion in healthy men and women at different ages. Eur J Appl Physiol. 88(3): 247–254.
van der Horst M. (2002) Human Head Neck Response in Frontal, Lateral, and Rear End Impact Loading – Modelling and Validation, Doctoral Thesis, Eindhoven University of Technology, Eindhoven, the Netherlands.
Vasavada A, Brault J, Siegmund G. (2007) Musculotendon and Fascicle Strains in Anterior and Posterior Muscles during Whiplash Injury. Spine 32(7): 756–765.
Vasavada A, Danaraj J, Siegmund G. (2008) Head and neck anthropometry, vertebral geometry and neck strength in height-matched men and women. J Biomech. 41(1): 114-121.
Vasavada A, Li S, Delp S. (2001) Three-Dimensional Isometric Strength of Neck Muscles in Humans. Spine. 26(17): 1904-1909.
Vavalle N, Moreno D, Rhyne A, Stitzel J, Gayzik F. (2013) Lateral Impact Validation of a Geometrically Accurate Full Body Finite Element Model for Injury Prediction, Ann Biomed Eng. 41(3): 497–512.
Watanabe Y, Ichikawa H, Kayama O, Ono, K, Kaneoka K, Inami S (2000) Influence of Seat Characteristics on Occupant Motion in Low-Velocity Rear-End Impacts, Accid Anal Prev 32(2): 243–250.
Watanabe R, Miyazaki H, Kitagawa Y, Yasuki T. (2011) Research of Collision Speed Dependency of Pedestrian Head and Chest Injuries Using Human FE Model (THUMS Version 4). 22nd ESV Conf. (Wasington D.C., June 13–16). Paper no. 11–0043.
Wheeldon J, Stemper B, Yoganandan N, Pintar F. (2008) Validation of a Finite Element Model of the Young Normal Lower Cervical Spine, Ann Biomed Eng. 36(9): 1458–1469.
Wild CY, Steele JR, Bridget J, Munro BJ (2012) Why Do Girls Sustain More Anterior Cruciate Ligament Injuries Than Boys? A Review of the Changes in Estrogen and Musculoskeletal Structure and Function During Puberty. Sports Med. 42(9): 733–749.
Yang K, King A. (2003) Neck Kinematics in Rear-end Impacts. Pain Res Manag. 8(2): 79–85.
Yang K, Zhu F, Luan F, Zhao L, Begeman, P. (1998) Development of a Finite Element Model of the Human Neck. In Proceedings of the 42nd Stapp Car Crash Conference (Tempe, AZ, USA). Paper no. 983157.
Yoganandan N, Pintar F, Gennarelli T, Eppinger R, Voo L.(1999) Geometrical Effects on the Mechanism of Cervical Spine Injury due to Head Impact. Proceedings of IRCOBI Conference, Sitges, Spain.
Yoganandan N, Sances A Jr, Maiman DJ, Myklebust JB, Pech P, Larson SJ (1986) Experimental Spinal Injuries with Vertical Impact. Spine. 11(9): 855–860.
Zhang Q, Teo E, Ng H. (2005) Development and Validation of a C0–C7 FE Complex for Biomechanical Study, J Biomech Eng. 127(5): 729–735.
Zhang Q, Teo E, Ng H, Lee V. (2006) Finite Element Analysis of a Moment-Rotation Relationship for Human Cervical Spine, J Biomech., 39(1): 189–193.