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Review

Biomechanical musculoskeletal models of the cervical spine: A systematic literature review

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ABSTRACT

Background: As the work load has been shifting from heavy manufacturing to office work, neck disorders are increasing. However, most of the current cervical spine biomechanical models were created to simulate crash situations. Therefore, the biomechanics of cervical spine during daily living and occupational activities remain unknown. In this effort, cervical spine biomechanical models were systematically reviewed based upon different features including approach, biomechanical properties, and validation methods.
Methods: The objective of this review was to systematically categorize cervical spine models and compare the underlying logic in order to identify voids in the literature.
Findings: Twenty-two models met our selection criteria and revealed several trends: 1) The multi-body dynamics modeling approach, equipped for simulating impact situations were the most common technique; 2) Straight muscle lines of action, inverse dynamic/optimization muscle force calculation, Hill-type muscle model with only active component were typically used in the majority of neck models; and 3) Several models have attempted to validate their results by comparing their approach with previous studies, but mostly were unable to provide task-specific validation.
Interpretation: EMG-driven dynamic model for simulating occupational activities, with accurate muscle geometry and force representation, and person- or task-specific validation of the model would be necessary to improve model fidelity.

1. Introduction

Neck pain is one of the three most commonly reported complaints of the musculoskeletal system (Trinh et al., 2006). It is estimated that the United States spends around $88B per year in direct costs treating patients with low back and neck pain, which is more than the amount spent on treating any other condition save for diabetes and ischemic heart disease (Dieleman et al., 2016).

The economies of the industrial world have shifted in that they were dependent on manufacturing but now rely largely on the service sector. The shift has transformed the nature of work injuries and disability. The high rate of acute and fatal injuries observed at the beginning of the 20th century has been replaced by a sharp increase in the incidence of work-related musculoskeletal disorders such as neck pain (Côté et al., 2008; Vasavada et al., 2015). While low back pain has traditionally been the most common spine-related complaint, more and more patients are presenting with neck and radiating arm pain. Annual prevalence rates for neck pain have grown to 27–48% and are expected to continue to rise due to growing sedentary life and work style (Côté et al., 2008, 2009). Preliminary evidence shows that occupation and occupational class is highly associated with the risk of neck pain. For instance, according to Côté et al., 2008, among health care workers, the annual prevalence of neck pain ranged from 17% in dentists, 26% in pharmacists and 72% in dental hygienists.

Strong associations have been observed between cervical spine myofascial pain to neurological and biomechanical interactions of muscles and neck posture (Hong et al., 2019). The range of motion of the neck and the activities of the cervical muscles appeared to be altered in myofascial patients. It has been demonstrated that majority of work-related neck disorders can be caused by head positioning

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disorders and dysfunctional neuromuscular control of neck muscles as a result of prolonged sitting at work (Kocur et al., 2018).

Prolonged, flexed cervical spine posture has been flagged as a major risk factor for the development of chronic neck pain (Ariëns et al., 2000). A compelling biomechanical mechanism relating repeated neck flexion to the initiation and presentation of cervical spine disc herniation was presented by (Tampier et al., 2007). Because of this plausible pathway leading from repeated flexion to chronic neck pain, the question remains how adopting posture effects the cervical spine compression and shear loads. Unfortunately, most patients who experience neck pain do not gain complete remission of their symptoms and disability (Côté et al., 2004; Vasseljen et al., 2013). Therefore, it is very important to consider prevention strategies for neck pain, considering the economic burden it poses and the significant impact on the quality of life of those suffering from it.

Despite the prevalence of neck disorders among working populations, the underlying mechanism explaining the cause of neck pain remains unclear. Cervical spine musculature plays an important role in maintaining head and neck stability and preventing intervertebral joint disorders (Van den Abbeele et al., 2018). Thus, abnormal cervical spine muscle behavior may be a potential pathway for the etiology of neck pain and cervical spine disorders (Cheng et al., 2014). Furthermore, the consequences of surgical complications are not yet fully understood, particularly adjacent segment disease and proximal junctional kyphosis (PJK), which might induce a secondary complication in the cervical spine. Abnormal spine loading and muscular dysfunction could also be a problem. Therefore, quantifying the spinal muscle force distribution and its effect on intervertebral joints during dynamic activities could provide valuable information to help guide clinical evaluation.

Due to its complex musculoskeletal structure, the human cervical spine is one of the most challenging areas for developing biomechanical modeling. According to (Panjabi, 1998), there are four major biomechanical modeling techniques including: physical models, in vitro models, in vivo models, and computer models. Physical models have been successfully used for fatigue test of various spine implants and instruments since they are relatively inexpensive and simple to use. However, as these models usually made of non-biological material, they are less reliable in compare to other models if the bony anatomy and biomechanical properties of soft tissues are important. Animal and human cadaver in vitro models are useful in providing general knowledge of the spine construct where the anatomy and biomechanical properties is important. However, these models are expensive, variable, and very difficult to obtain. In vivo biomechanical models are generally animal models, and rarely human volunteers. These models have been used to study a living phenomenon such as degenerative process, fusion, and soft tissue injury. Considering the differences in anatomy and biology of different animal species and human, such models are less reliable, considering the human in vivo models cannot be used to evaluate injury mechanism, and anatomical and biological differences between human and animals. Computer (numerical) models on the other hand, are a set of mathematical equations that incorporate both the geometry, physical, and biomechanical properties of the represented structure. Certain advantages inherent in computer models make them more appealing for assessing underlying injury mechanisms, and to investigate biomechanical parameters that cannot be measured directly through in-vitro or in-vivo tests or in hazardous conditions that ethically are not feasible (Ahn, 2005; Cazzola et al., 2017). Since these models are not universal, there are certain concerns associated to them including challenging validation process, and the application of the model beyond its validation boundaries.

A wide array of computational cervical spine models exist within the scientific literature, but they have yet to be compared to one another to provide guidance on the current needs for improvement. Therefore, the objective of the current literature review was to evaluate the performance and structural characteristics of existing computer head and neck models in the literature. This review is intended to provide a better understanding of current state-of-the-art and help inform future developmental efforts toward more accurate and realistic cervical spine models for occupational neck injury risk evaluation.

2. Methods

A systematic literature review was performed through PubMed, Web of Science, and Science Direct. The selection criteria considered searched for studies including the following: (1) ‘cervical’, and (2) ‘spine’ or ‘neck’, and (3) ‘model’ or ‘biomechanical’. Additional criteria included language in ‘English’ and a study population of ‘humans’. Articles cited in all retrieved studies were also searched to gather additional sources.

The primary search results were screened initially by their title, then abstract, and finally full text to meet several restricted criteria for this study. First, the study must have included a biomechanical model. Second, the study must have represented entire cervical spine Skull-C7. Third, the study must have represented cervical spine muscles as separate structural elements rather than lumped parameters within intervertebral joint stiffness. Fig. 1 illustrates literature search and selection process.

Selected studies first were incorporated in one of the three categories based on their biomechanical modeling approach: (1) multibody dynamic model; (2) finite element model; (3) hybrid model. Then, they were further categorized by structural modeling characteristics: (1) analysis type ‘dynamic’ or ‘static’; (2) muscle lines of action ‘straight’ or ‘curved’; (3) muscle curvature method ‘via-point’ or ‘wrapping surface’; (4) muscle force model ‘active’ and/or ‘passive’; (5) muscle activation dynamic ‘pre-defined’, ‘optimization’ or ‘EMG-driven’; (6) personalized muscle parameters (geometry and force). Then within each model the performing task, validation method, and model performance evaluation was recorded.

Structural modeling characteristics, (1) ‘dynamic’ or ‘static’ term refers to the capability of the biomechanical model to simulate static or dynamic load; (2) ‘straight’ or ‘curved’ defines whether muscle geometry is represented by a straight line connecting origin to insertion or as a curve line/surface; (3) ‘via-point’ or ‘wrapping surface’ specifies whether intersegmental points (via-points) have been used or a wrapping surface is incorporated to account for muscle curvature; (4) ‘active’ and ‘passive’ identifies muscle force model's component; (5) ‘pre-defined’, ‘optimization’ or ‘EMG-driven’ term refers to whether muscles activation dynamic within the model were derived from pre-defined active state curve, estimated using optimization algorithms or is set based on direct measurements such as EMG signals, (6) “Personalized” muscles refer to considering individual differences in muscle morphological and physical characteristics (CSA, length, moment arm, etc.) and force generating potentials (muscle F-L and F−V relationships, and muscle strength effecting active and passive force).

3. Results

The initial search using the aforementioned search terms led to a total of 248 articles in Web of Science, Science Direct, and PubMed. Screening of titles, abstracts, and full text for each initially retrieved article was performed based on the strict criteria of this study as described in method. A total of 58 studies met the selection criteria and were included in this review. In total, 22 different cervical spine models were recognized within the reviewed articles. Tables 1 and 2, summarize modeling techniques and model structure properties within the identified models, respectively. (See Fig. 1.)

3.1. Model approach

Within the reviewed studies, the ‘multibody dynamic’ method was used in 13 models, the ‘finite element’ method was implemented in two studies, and hybrid (multibody and finite element) method was used in

Table 1
Cervical spine computer models within retrieved studies; model type and validation technique.

#	Study	Model type	Validation task	Validation technique	Performance measure
1	(Brolin, 2002; Brolin et al., 2005, 2008; Brolin and Halldin, 2004)	FEdynamic	Frontal and lateral impact	Head kinematics compared to human volunteer sled test (Ewing et al., 1976, 1978)	Correlation (not specified)
2	(Meyer et al., 2004, 2013)	FE/dynamic	Front, lateral and rear impact	Head and neck kinematics compared to human volunteer sled test (Ewing et al., 1976, 1978)	Correlation coefficient (R^2)
3	(de Jager, 1996; de Jager et al., 1996, n.d.)	HB/dynamic	Front and lateral impact	Head kinematics compared to human volunteer sled test (Ewing et al., 1976, 1978)	Qualitative-visual inspection
4	(Van Ee et al., 2000)	HB/dynamic	Quasi-static tension	Location of injury compared to cadaveric experiment (Van Ee et al., 2000)	RMSE
5	(Deng and Fu, 2002)	HB/dynamic	Frontal impact	Head kinematics compared to human volunteer sled test (Ewing et al., 1976, 1978)	Correlation coefficient
6	(Chancey et al., 2003)	HB/dynamic	Quasi-static tension	Location of injury compared to cadaveric experiment (Van Ee et al., 2000)	Quantitative comparison between spinal loads
7	(Esat and Acar, 2009; Lopik and Acar, 2004; van Lopik and Acar, 2007)	HB/dynamic	1. Single plane motion 2. Rear impact	Head and neck kinematics compared to (Grauer et al., 1997)	Qualitative comparison
8	(Fice et al., 2011; Panzer et al., 2011)	HB/dynamic	Front and rear impact	1. Ligament strain compared to experimental (Ivancic et al., 2004; Pearson et al., 2004) 2. Head kinematics compared to human volunteer test (Davidsson et al., 2001; Deng et al., 2000)	Within ± standard deviation
9	(Dibb, 2011; Dibb et al., 2013; Nightingale et al., 2016)	HB/dynamic	1. Muscle path validation 2. Dynamic flexion/extension 3. Frontal impact	1. Compared straight line with/without via-point and wrapping surface with centroid path derived from MRI 2. Kinematic validation for dynamic flexion/extension (Wheeldon et al., 2006)	1. Average absolute error 2. Correlation coefficient (R^2)
10	(Deng and Goldsmith, 1987; Merrill et al., 1984)	MB/dynamic	Frontal and lateral impact	1. Head kinematics compared to human volunteer sled test (Ewing et al., 1976, 1978) 2. Muscle strain compared to human volunteer sled test (Ewing et al., 1976, 1978)	Qualitative-visual inspection
11	(Moroney et al., 1988a)	MB/quasi-static	MVC: flexion, extension, lateral bending and axial rotation	Mean of the model prediction muscle forces with mean sEMG signals corresponding to the same muscle group	Correlation coefficient (R^2)
12	(Snijders et al., 1991)	MB/static	None	None	None
13	(Horst and Der, 2002; Horst et al., 1997)	MB	1. Isometric flexion/extension 2. Dynamic flexion/extension 3. Frontal and lateral impact	1. Maximum moment about C7 was compared with reported experiment measurements from isometric experiment(Snyder et al., 1975) 2. Intervertebral joint kinematics in static spine loading was compared to experimental tests (Moroney et al., 1988b) 3. Intervertebral joint flexibility during neck flexion/extension compared to experimental tests (Camacho et al., 1997) 4. Head and neck kinematics compared to human volunteer sled test (Ewing et al., 1976, 1978)	Qualitative comparison
14	(Vasavada et al., 1998, 2007, 2008)	MB/isometric	1. Single plane motion 2. Rear impact	1. Muscle straight LOA was compared to centroid path obtained from MRI 2. Muscle strain vs muscle activity recorded by surface and intramuscular EMG	Average absolute error Qualitative comparison
15	(Choi, 2003; Choi and Vanderby, 2000; Choi and Vanderby Jr, 1999)	MB/quasi-static	MVC: flexion, extension, lateral bending and axial rotation	Predicted muscle moments versus measured moments about C4–C5 centroid	RMSE
16	(Yamazaki et al., 2000)	MB/dynamic	Rear impact	Intervertebral joint angles compared to experiment (ONO and KANEOKA, 1999)	Qualitative comparison
17	(Kruidhof and Pandy, 2006; Oi et al., 2004)	MB/dynamic	MVC: flexion, extension and bending	Muscles moments compared with literature (Peolsson et al., 2001; Seng et al., 2002)	Error
18	(Netto et al., 2008)	MB/dynamic	Isometric flexion/extension	Muscle moments compared to externally measured moments by dynamometer	Correlation coefficient (R^2)
19	(Suderman et al., 2012; Suderman and Vasavada, 2013, 2017)	MB/static	Sagittal plane posture	Muscle straight LOA was compared to centroid path obtained from MRI	Error
20	(Huber, 2013)	MB/dynamic	Flexion, lateral bending	Comparison between model predicted moments from muscles versus measured moments by force plate and motion capture	Correlation coefficient (R^2) Average absolute error
21	(Cazzola et al., 2017)	MB/dynamic	1.Single plane motion(flexion, extension, lateral bending and axial rotation) 2.Front impact	1. Kinematic validation (Ludewig et al., 2009) 2. Muscle excitation pattern calculated within optimization algorithm compared to recorded sEMG	Qualitative comparison
22	(Mortensen et al., 2018)	MB/dynamic	Single plane motion(flexion, extension, lateral bending and axial rotation)	Muscle moments compared to externally measured moments (Fice et al., 2014)	Error

7 models.

3.2. Model structural characteristics

3.2.1. Static/dynamic

16 models were able to simulate dynamic conditions in that head and vertebral bodies linear and angular velocity and acceleration were taken into account.

3.2.2. Muscle lines of action (LOA) 'straight' or 'curved'

Results showed that straight LOA was used in all of the 22 models. In addition, two models also incorporated both straight and curve LOA to investigate effects of straight versus curved muscle LOA on model performance.

3.2.3. Via-point/wrapping

Among all models with straight LOA muscle representation obtained by connecting muscle origin to insertion in supine or neutral standing postures, 10 models incorporated frictionless via-points to consider muscle curvature around the spine in dynamic motions. Among these models, two models developed straight LOA with/without via-point, as well as a wrapping surface for some muscles in order to evaluate the accuracy of these different approaches.

3.2.4. Muscle force model

Hill-type contractile element of muscle force was ultimately implemented in 18 models. 15 models used a non-linear stress-strain curve to account for muscle passive force. Only 3 studies implemented a Hill-type passive element.

3.2.5. Muscle active state pre-defined/optimization/EMG-based

A pre-defined activation-time curve was implemented in 14 models in order to allow forward dynamic simulation. Four models used an inverse dynamics optimization approach to predict muscle forces and subsequently, muscle activity level. Only three models used EMG signals as input for muscle active force calculation.

3.2.6. Personalized muscle

A generic cervical spine skeleton and muscle geometry using previous literature data were most commonly used (15 models). Only 7 models were scaled to match the musculoskeletal geometry of the subject's anthropometric measures. All the models used generic values for at least one of the muscle force parameters such as cross-sectional area (PCSA), maximum isometric stress, optimum muscle length, active and passive force-length curve shape, or maximum shortening velocity. No models fully personalized the cervical spine skeleton, muscle geometry, and muscle force parameters.

3.2.7. Spine passive elements

Intervertebral disc stiffness, ligament force, and facet contact forces were not represented in 7 models. In 8 models, each of these structures were modeled as independent elements, while 5 models used a lumped parameter for intervertebral joint stiffness instead of individual elements.

3.3. Validation

3.3.1. Validation task

Among 22 models studied in this literature review, 10 models were developed to investigate impact situations (front, rear and lateral) such as automotive crash tests or athletic engagement. A small number of studies (4 articles) considered basic exertions such as main physiological plane movements (flexion, extension, lateral bending) or upright posture; five models were validated for isometric conditions.

3.3.2. Validation technique

Table 1 shows the implemented validation strategy for each model. Five models were validated with multiple independent variable measurements. Kinematic validation was the most common technique used. Validating muscles LOA were performed for two models. Muscle generated moments compared to externally measured moments were used as a validation technique in only 4 models. Investigating model performance to accurately predict muscle force, compared to recorded EMG signals was used in two models.

3.3.3. Performance measure

Qualitative comparison was the most common assessment (7 models) method followed by correlation coefficient and average error.

4. Discussion

4.1. Overall observation

The presented literature review examined biomechanical models of the cervical spine. Based on the systematic evaluation, the most common cervical spine computer modeling approaches are finite element (FE), multibody dynamic (MB), and hybrid (HB) methods.

Finite element models, which are constructed of many small elements, have been used in biomechanical analyses to study the stresses of tissues under external loads. Finite element models are able to represent complex geometries and simulate material behavior in detail. Multi-body dynamic models are mathematical models composed of rigid bodies representing bone and spring-damper elements representing interconnecting soft tissue. Mathematical equations representing the behavior of spring-damper elements govern the kinematic and kinetic response of the models to externally applied forces.

With the intention of overcoming the limitations of the existing dynamic biomechanical model and traditional FE, the hybrid FE-MB models have been further developed to include much greater anatomic detail and to take advantage of flexible multi-body. Hybrid models greatly reduces an FE model's computational complexity with a minimum loss of accuracy while allowing large overall motion and complex interaction with other elements. Ultimately, this allows complex structures, such as the intervertebral disc, to be accurately represented as a flexible body in a dynamic model.

Among reviewed models, straight line muscle models with generic muscle morphological properties, forward-dynamic models with predefined muscle activation mechanics, and impact scenario evaluations were the most common. In comparison to other joints of the body, the cervical spine is more complex to model, due to its anatomical complexity. The complex interaction between atlas and the axis, along with several ligaments and muscles specialized for that region, enabled those joints to dominant head and neck total range –of-motion without compensating stability of the region. To model such complex joints, and intricate musculoskeletal orientation, researchers had to implement significant amounts of simplifications and assumptions during their developmental efforts (Netto et al., 2008). Therefore, it is important to note that the most common modeling approach and characteristics among cervical spine models that is currently being used among many researchers, doesn't certify their validity (Hwang et al., 2017).

4.2. Model critique

Multibody computer models of the head and neck are composed of rigid bodies connected via joints, usually consisting of springs and dampers. More complex MB models represent the head, cervical vertebrae, and torso as rigid bodies connected with joints which represent the intervertebral discs, cervical ligaments, and facet joints. The FE musculoskeletal models of the head and neck on the other hand, mostly include deformable vertebrae and a head connected through deformable intervertebral discs, cervical ligaments, and facet joints modeled

Table 2
Cervical spine computer models within retrieved studies; muscle modeling technique.

#	Study	Muscle geometry		Muscle force		Muscle active state	Personalized muscle		F-L/F-V		Spine passive elements
		Straight/curve	Via-point/wrapping surface	Active	Passive	Pre-defined/optimization/EMG-driven	Geometry	Force	F-L	F-V	
1	(Brolin, 2002; Brolin et al., 2005, 2008; Brolin and Halldin, 2004)	Straight	4 spring elements in series for superficial muscles	Hill-element	Nonlinear spring	Pre-defined Activation-time curve	No	No	No	No	Yes Individual elements
2	(Meyer et al., 2004, 2013)	Straight and curve	None	Hill-element	Continuum mechanics	Pre-defined Activation-time curve	Yes	No	No	No	Yes Individual elements
3	(de Jager, 1996; de Jager et al., 1996, n.d.)	Straight	None	Hill-element	Nonlinear stress-strain curve	Pre-defined Activation-time curve	No	No	No	No	Yes Lumped behavior
4	(Van Ee et al., 2000)	Straight	None	No	Nonlinear stress-strain curve	Pre-defined Activation-time curve	No	No	No	No	Yes Lumped behavior
5	(Deng and Fu, 2002)	Straight	1 via-point	Hill-element	Hill-element	Pre-defined Activation-time curve	No	No	No	No	Yes Individual elements
6	(Chancey et al., 2003)	Straight	None	Hill-element	Nonlinear stress-strain curve	Optimization Minimizing muscle fatigue	No	No	No	No	Yes Lumped behavior
7	(Lopik and Acar, 2004; van Lopik and Acar, 2007)	Straight	Multiple via-point	Hill-element	Nonlinear stress-strain curve	Pre-defined Activation-time curve	No	No	Yes	Yes	Yes Individual elements
8	(Fice et al., 2011; Panzer et al., 2011)	Straight	None	Hill-element	Hill-element	Pre-defined Activation-time curve	No	No	No	No	Yes Individual elements
9	(Dibb, 2011; Dibb et al., 2013; Nightingale et al., 2016)	Straight	Multiple via-point	Hill-element	Nonlinear stress-strain curve	Pre-defined Activation-time curve	No	No	No	No	Yes Lumped behavior
10	(Deng and Goldsmith, 1987; Merrill et al., 1984)	Straight	1 via-point	None	Non-linear spring	Pre-defined Activation-time curve	No	No	No	No	Yes Lumped behavior
11	(Moroney et al., 1988a)	Straight	None	Hill-element	None	Optimization Minimum contraction intensity	Scaled	No	No	No	No
12	(Snijders et al., 1991)	Straight	None	Muscle force	Optimization Minimizing net joint moment	No	No	No	No	No	No
13	(Horst et al., 1997; Horst, 2002)	Straight	Multiple via-point	Hill-element	Nonlinear stress-strain curve	Pre-defined Activation-time curve	No	No	Yes	Yes	Yes Individual elements
14	(Vasavada et al., 1998, 2007, 2008)	Straight	Multiple via-point	Hill-element	Nonlinear stress-strain curve	Pre-defined Activation-time curve	Scaled	No	Yes	Yes	No
15	(Choi, 2003; Choi and Vanderby, 2000; Choi and Vanderby Jr, 1999)	Straight	None	Hill-element	None	EMG-assisted Optimization Moment matching	Scaled	No	No	No	No
16	(Yamazaki et al., 2000)	Straight	None	None	Nonlinear stress-strain curve	Pre-defined Activation-time curve	No	No	No	No	Yes Individual elements
17	(Kruidhof and Pandy, 2006; Oi et al., 2004)	Straight	Multiple via-point	Hill-element	Nonlinear stress-strain curve	Pre-defined Activation-time curve	Scaled	No	Yes	Yes	Yes Only ligaments
18	(Netto et al., 2008)	Straight	Multiple via-point	Hill-element	Nonlinear stress-strain curve	EMG	Yes	No	Yes	Yes	No
19	(Suderman et al., 2012; Suderman and Vasavada, 2013, 2017)	Straight and curve	Multiple via-point /wrapping surface	Hill-element	Nonlinear stress-strain curve	Pre-defined Activation-time curve	Scaled	No	No	No	No
20	(Huber, 2013)	Straight	None	Hill-element	Hill-element	EMG	No	No	No	No	Yes Individual elements
21	(Cazzola et al., 2017)	Straight	None	Hill-element	Nonlinear stress-strain curve	Optimization Minimizing sum of the square of muscle activations	Scaled	Scaled	No	No	No
22	(Mortensen et al., 2018)	Straight	None	Hill-element	Nonlinear stress-strain curve	Optimization Minimizing sum of the square of muscle activations	Scaled	Scaled	Yes	Yes	Yes Lumped behavior

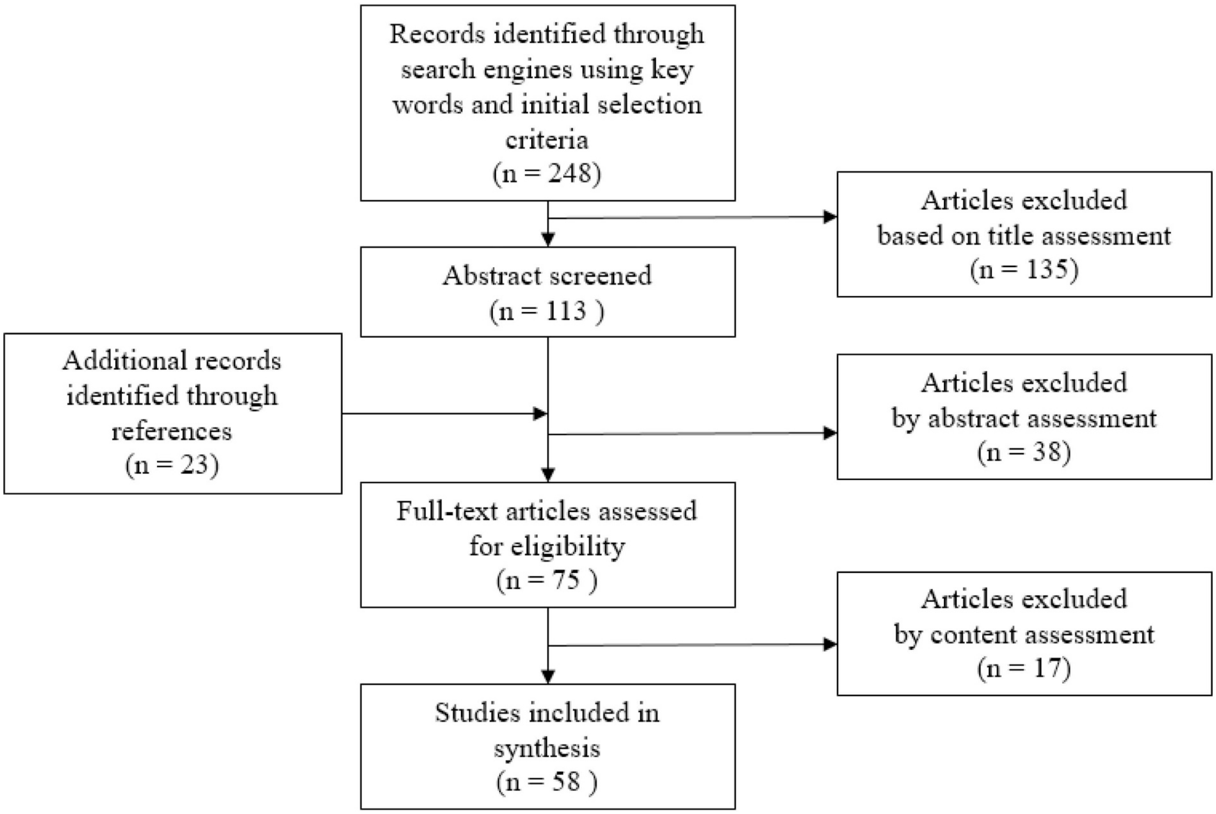

Fig. 1. Flow chart illustrating literature search and selection process.

by mechanical elements. The FE models have capability to model intervertebral disc and soft tissues in great detail and to provide more visual tool to understand injury mechanism and site of injury within specific tissue which advanced them from MB models (Ahn, 2005). The MB models have several advantages over FE models, the most important being that they are computationally efficient. In addition, FE models are mostly used for simulating static or isometric situations, while multi body simulations are capable of simulating high velocity dynamic motions. The great computational demands inherent in FE models usually restrict models to only small portions of the spine and either static or short duration quasi-dynamic analyses. These models also usually simplify the representation of the spine musculature. The lack of realistic muscle representation and dynamic loading limits FE model's utility in accurately representing the entire cervical spine and its response to various occupational setting (Ahn, 2005). Therefore, for investigating cervical spine injury risk associated to occupational environments or impacts, multi body dynamic simulation appears to be more appropriate.

4.2.1. Muscle representation

Biomechanical models of the cervical spine must include biofidelic representation of musculature, as they are an important structural components of the neck (Dibb, 2011). Researchers have shown muscles in cervical spine models not only provide stability under gravitational forces (Dibb et al., 2013) but affect the magnitude and timing of peak spinal loads (Nightingale et al., 2016). Based on this review, 'straight' LOA was the most common technique used in cervical spine models. In this method, the muscle path is generally defined as a straight line connecting the muscle origin to its insertion. These single segment muscles are only able to interact at their connected endpoints. If a muscle spans several cervical vertebrae, the muscle lacks the ability to interact with those spanned vertebrae. Straight LOA models respond reasonably well in single-plane tasks. However, this assumption is unreliable for more complex occupational tasks commonly required in working environments (Hwang et al., 2016). Realistic representation of muscle LOA in biomechanical computational models is a key parameter for accurately estimating muscle moment generating capacity and consequently spinal load (Vasavada et al., 2008).

Alternatively, curved muscle paths including multi-segmented muscles and wrapping surfaces using shell or solid elements have been suggested. A multi-segmented method refers to discrete segments of muscle elements at every level that the muscle has attachments. The intersegmental nodes are attached to adjacent vertebra at a fixed distance. According to Suderman and Vasavada (2012), moving intersegmental nodes to account for muscle shape variation with postural changes which are not considered with fixed points set at a neutral posture. A wrapping surface, on the other hand, is a geometric surface constraint where line segments wrap over it. Such a surface is constraining the muscle LOA to pass it and instead to wrap around it on depending on the posture. Considering the complexity of the cervical spine musculoskeletal system, it is very challenging to develop a wrapping surface for all muscles that accurately represents various anatomical postures as compared to structural joints such as knee or elbow, where wrapping surfaces have been used very frequently. Regarding model performance, the multi-segmented method with fixed intersegmental points showed significantly lower deviation from the standard muscle centroid path obtained from MRI in multiple neck postures when compared to the wrapping surface method (Suderman et al., 2012; Suderman and Vasavada, 2017). Given this, it seems reasonable to develop multi-segmental muscle path with via points for cervical spine models.

Among the reviewed studies, a Hill-type muscle model was used exclusively in different capacities. The Hill-type muscle model is a phenomenological model inheriting from early work of (Winters, 1990). In this model, the muscle is partitioned into a contractile element representing muscle active behavior, a series elastic element representing tendons, and a parallel elastic element representing passive muscle behavior. Some models are only capable of simulating passive muscle behavior by means of nonlinear force/stress-strain relationship. In a study performed by Horst and Der (2002), comparing experimental data and simulation results, indicated neck muscles can alter head and neck kinematics. They found varying muscle activation showed great influences on the internal loads for the intervertebral discs, facet joints and ligaments. They had concluded, ignoring active muscle force in such models could drastically under predict spinal loading by ignoring the contractile element potential for generating force. Therefore, it is essential for a valid model to embed both contractile and passive elements of the Hill-type muscle model.

Realistic computational models of the head and neck have to include accurate representations of the cervical musculature and other significant passive components of the neck such as intervertebral discs, cervical spine ligaments, and facet contact forces (Dibb, 2011). Most of the current models used lumped parameter intervertebral joint to account for all of the passive components on the joint. Such assumptions may drastically affect both model kinematics and predicted spinal loads. These models are limited because excluding physical models of ligaments and facet contacts ignores the effect of passive force loading on the vertebral bodies. Therefore, it is important to model head and neck passive components as individual structures.

Some of the cervical spine musculoskeletal models were scaled for skeleton dimensions, but this technique does not take into account person-specific muscular geometry and muscle force characteristics, which are fundamental to represent specific pathologies in clinical applications (Klein Horsman et al., 2007) and to describe the morphology of specific populations (Cazzola et al., 2017). Since Vasavada et al., 1998 showed the model has been highly sensitive to variations in muscle PCSA, there is high probability that incorporating subject-specific muscle parameters would have a significant influence. Therefore, we believe muscle-specific and subject-specific parameters such as gain will significantly improve cervical spine model's performance toward accuracy. The force-length and force-velocity relationships in combination with the muscle active state constitute Hill-type contractile muscle model. Most of the models do not account for force-length and force-velocity factors when estimating muscle force (Table 2). Others assume the same relationship for all subjects. Previous studies have shown muscle force-length and force-velocity are related to age (Thelen, 2003) and optimal sarcomere length variation among human subjects (Lieber et al., 1994; Walker and Schrodt, 1974). Various shape factors of muscle force-length and force-velocity relationships directly influence the magnitude and chronological variability of muscle forces as a function of muscle length and velocity changes among individuals. Mortensen et al., 2018 found that person-specific model parameters affect model spinal load predictions. Considering the physiological variability of muscle properties could help to estimate more accurate muscle forces. Therefore, it is critical to develop a person-specific musculoskeletal model to account for individual muscle physiological and morphological differences.

Recently, a possible mechanism was proposed which links chronic neck pain to maladaptive muscle activity observed in the upper trapezius and scalenius (Falla et al., 2004, 2007). In addition, reported abnormal muscle activity in an agonist muscle group and an increase in co-contraction in superficial muscles, with a significant decrease in activation of the deep cervical spine muscles, is associated with neck pain (Jull, 2000). Therefore, it is critical to estimate accurate muscle active states while accounting for co-contraction among agonist and antagonist muscles for correct muscle force calculation and injury risk evaluation. The forward dynamic and optimization methods were the most common methods used to estimate muscle force (and consequently active state) across the studies. In the forward dynamic method, muscle activation is set by a pre-defined activation curve. Using these non-physiologic techniques, no study to date has been able to provide a realistic set of activations to maintain the head in an equilibrium at upright posture under kinematic effect of gravity to run impact simulations (Chancey et al., 2003). Moreover, to date no model has been able to create a set of activations that represents subjects in a crash (de Jager et al., 1996; Horst et al., 1997). On the other hand, the optimization based models calculate reaction forces and moments about the joint based on known or measured external forces. Then, muscle forces and moments are calculated by application of an objective function. This approach has been criticized for several reasons. First, it usually does not predict co-contraction of antagonist muscles (Choi and Vanderby Jr, 1999; Netto et al., 2008), which is seen in complex dynamic motions. Studies had shown that ignoring co-activation could result in underestimating spinal load 45%–70% (Granta and Marras, 1999). Choi (2003) showed that co-contractions are essential to provide stability in the human cervical spine. Second, some discrepancies between model predictions and experimental measures were seen. For instance, Moroney et al. (1988a) reported coefficient of correlation between estimated muscle force and measured were mostly less than 0.5. Third, predicted muscle forces via the optimization approach greatly depend on the defined objective function and sets of constraints (Choi, 2003; Mortensen et al., 2018). Choi and Vanderby Jr, 1999 showed how different optimization algorithms generate different muscle forces. On the other hand, an EMG-based modeling approach relies on measured EMG signals to calculate muscle forces with accurate determination of antagonistic muscles co-contractions together with agonistic synergy.

The EMG-based modeling approach accounts for subject and trial differences in the magnitudes of individual muscle forces needed to perform the task. While in forward dynamic and optimization methods similar estimates of muscle forces for all subjects and performed tasks was seen (Choi and Vanderby, 2000). However, one of the limitations associated to EMG-based modeling technique is the inaccuracy of perceiving signals from deep muscles by surface EMG. To consider deep muscle moment generating potential, an EMG-assisted optimization method has been introduced. EMG-assisted optimization method implements direct measurements to drive muscle activity. Then, within an optimization algorithm, personalization within the model taken placed by predicting parameters such as muscle gain, active and passive force-length and force-velocity (Choi, 2003; Choi and Vanderby, 2000; Choi and Vanderby Jr, 1999). While these studies are useful in understanding optimization, due to the poor neck muscles representation as static forces in single level and their inability to model dynamic of whole head and neck, this model have had limited predictive ability. In addition, a nonlinear EMG-force relationship is suggested to be present in modeling cervical spine EMG-driven models. This is based on the suggestion by (Buchanan et al., 2004) that EMG input in biomechanical models should be further investigated with consideration of a nonlinear EMG-force relationship.

4.2.2. Kinematics

Most of the available models in the literature were developed to investigate the precise mechanism of cervical spine injuries during sports and traffic collisions. Although such models are capable of simulating dynamic motions, their structural characteristic prevent them from performing well for complex dynamic daily-living activities. Therefore, in order to biomechanically investigate causal pathways of neck pain among the working population, a cervical spine model capable of accurately simulating complex dynamic daily-living activities is needed. Biofidelic musculoskeletal models of the cervical spine with this ability will aid in understanding the role of neck muscles during motion, with applications in ergonomics, rehabilitation, and analysis of disease. Some models were claimed as capable of dynamic simulations, however they were only validated for static /quasi-static tasks. Most of the dynamic models had been validated for impact/crash conditions and only few models were validated for single plane motions. Within the scopes of the current literature review article, we haven't found any model validated for complex multi-plane dynamic motions. Such motions are very common in daily-living activities and are required in many occupational environments.

4.2.3. Validation

Developed cervical spine models have used different types of validation measures. Mostly, kinematic comparison between model predictions and experimental measurements were performed. Kinematic parameters have included head center of mass location, velocity and acceleration, and intervertebral joint ROM. Even though this method provides a reasonable insight into the performance of cervical spine as a whole structure, it doesn't offer any insight into individual soft tissues' contribution in total motion. For instance, inverse-dynamic models are capable of predicting different combinations for muscle forces to produce the same motion. These solutions depend on the defined objective function which often attempt to minimize muscle moments about the given joint. In this case, even if a model accurately predicts cervical spine kinematics, it is possible for spinal loads to be underestimated as a result of the lack of muscle co-contraction. Nevins et al., 2014 could quantify significant variation among healthy individuals in cervical spine intervertebral kinematics resulting from physiological variations. Therefore, it is unlikely that motion from a generic model would closely mimic subject specific kinematics. Some studies compared recorded surface or intermuscular EMG signals with model predictions. This method provides a good insight into the validity of model prediction for that specific task or posture. Spinal load and intervertebral disc stress were also evaluated in a few hybrid cervical spine models. These values were usually compared to the reported injury location during cadaver tests and measured intradiscal pressure. A few studies compared model estimation for muscle path with MRI-derived muscle centroid path in the same posture. Curved muscle path based on via-point technique showed insignificant difference when compared to curve muscle model based on wrapping surface. This measure was able to validate curve muscle models in certain postures such as neutral posture or single-plane flexion. However, this method has not been used to validate multi-plane postures and has only been used for a few muscles. The moment-generating capacity was calculated as the product of muscle moment arms and maximum isometric force and had been used as model validation when compared to experimentally measured neck muscle strength in certain positions. In general, for calculating the total moment-generating capacity, all surrounding muscles were assumed as maximally activated. Therefore, this assumption does not accurately represent realistic antagonistic muscles co-contraction. For this reason, this may not be reliable technique. Moment generating potential for complex dynamic motions was not investigated.

Moment-matching is considered the “gold-standard” validation method for model performance (Netto et al., 2008). Generally, an external moment about the joint is measured synchronously and compared to the model's internal moment prediction. A few studies had implemented the moment-matching validation technique for static trials and single-plane postures (Huber, 2013; Mortensen et al., 2018). The coefficient of determination (R^2) was used to validate the dynamic moment pattern between model prediction and measured external moments about the modeled joint. Average error (Nm) between estimated and measured moments was also used.

4.3. Summary

This review has helped identify several potential research directions for future cervical spine musculoskeletal models. First, realistic muscle LOA capable of accurately following spine curvature in dynamic motions have to be considered. Second, most of the existing models use generic musculoskeletal structures and muscle parameters. Future studies are recommended to develop predictive models of person-specific muscle paths and skeletons based on medical imaging such as CT, MRI and anthropometric measurements. Third, in an occupational environment and daily living activities, complex multi-planar motions are very common. Since it is known that such activities may increase risk of neck pain, more robust models capable of simulating such tasks is highly needed. Fourth, since most of the existing models implement pre-defined activation curves or optimization techniques to obtain muscle forces, they were unable to provide reliable feedback of muscle co-contraction and consequently spinal load. Therefore, an EMG-assisted cervical spine model is needed to better document spine loading associated with dynamic motions. Fifth, the existing models either lack force-length and force-velocity relationships for contractile muscle force calculations or consider generic properties for all the subjects. Further investigation on personalized muscle parameters are needed.

5. Conclusion

This literature review systematically summarized existing head and neck models with muscles modeled as independent structural elements. Multibody dynamics, followed by finite element and hybrid approaches were the most common modeling approach. Most of the existing models were designed to respond to impact conditions. Forward and inverse dynamics were frequently used for muscle force calculations, while EMG rarely was used. Hill-type muscle models were commonly used in different models. Some studies only used contractile elements, some only passive, and only a few models included both active and passive muscle force components. None of the models used complete Hill-type muscle models, as they mostly ignored force-length and force-velocity relationships. Another major limitation within existing models is generic muscle morphological properties such as cross-sectional area and muscle attachments, as opposed to person-specific data obtained from anthropometry measurements and medical imaging.

Future models should be developed that are capable of simulating complex dynamic activity of daily living and occupational tasks. EMG-assisted models are capable of accounting for muscle co-contraction and individual variations, with personalized musculoskeletal properties. Such a model should serve as a reliable platform to investigate occupational related risk factors for neck disorders.

Declaration of Competing Interest

None declared.

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