DRDC | RDDC
technologysciencetechnologie

The Use of Models in Aircrew Neck Trouble Research

Philip S. E. Farrell
DRDC – Toronto Research Centre

Barry S. Shender
Naval Air Warfare Center Aircraft Division

Giovanni Fusina
DRDC – Toronto Research Centre

Defence Research and Development Canada

Scientific Report
DRDC-RDDC-2018-R147
July 2018

Canada

IMPORTANT INFORMATIVE STATEMENTS

This document was reviewed for Controlled Goods by Defence Research and Development Canada (DRDC) using the Schedule to the *Defence Production Act*.

Endorsement statement: This publication has been peer-reviewed and published by the Editorial Office of Defence Research and Development Canada, an agency of the Department of National Defence of Canada. Inquiries can be sent to: Publications.DRDC-RDDC@drdc-rddc.gc.ca.

This work is a deliverable under 03aa Air Human Effectiveness Project and the Neck and Back Trouble Mitigation Solutions Work Breakdown Element.

Abstract

Seventy-five percent of Griffon Helicopter aircrew have reported various levels of neck pain as a result of performing flying tasks, and 23% have been grounded due to neck trouble. In the past, Defence Research and Development Canada (DRDC) has been involved in neck trouble research focussing on Finite Element modelling techniques in order to understand neck pain mechanisms. Some have argued that this line of research has not led to tangible solutions in Canada; however, others have used modelling as a necessary tool in the development of neck pain mitigating solutions. This paper argues that conceptual, physics based, and empirical models can be used at various stages to help define the scope of the problem, suitable verification and validation tests and provide a path for the solution's development, design, and assessment. This has led to a description of the NATO HFM 252 framework as well as mathematical equations that estimate neck loading due to helmet system mass properties. It is concluded that these modelling activities are essential for ethical and cost effective neck trouble research.

Significance to defence and security

This work is part of the Royal Canadian Air Force (RCAF) and DRDC Air Agile Program Air Human Effectiveness project (03aa) Neck and Back Trouble Mitigation Solutions Work Breakdown Element (WBE). It also has strong ties with the North Atlantic Treaty Organisation (NATO) Research and Technology Organisation (RTO) Human Factors and Medicine (HFM) Panel Research Task Group (RTG) HFM-252 on Aircrew Neck Pain as well as The Technical Cooperation Program (TTCP) Technical Panel (TP) 22 Military Medicine's Aircrew Spinal Pain Mitigation Collaborative Project (CP).

One of the intermediate outcomes as outlined in the Air Agile Program is to promote "Improved health and safety of operational RCAF personnel by exploring causes and solutions for aero-medical challenges such as hypoxia and neck strain injuries (DND/CAF, 2013)." This paper addresses how models and modelling can be used in the development and assessment of proposed solutions for neck trouble currently experienced by aircrew. This paper provides the proper context for modelling so that developers and assessors remain focused on the delivery of solutions, and not solely on modelling for understanding neck pain mechanisms.

Résumé

Soixante-quinze pour cent des membres d’équipage des hélicoptères Griffon ont signalé divers niveaux de douleur au cou à la suite de l'exécution de tâches en vol, et 23 % ont été cloués au sol en raison de maux de cou. Par le passé, Recherche et développement pour la défense Canada (RDDC) a participé à des recherches sur les maux de cou axées sur les techniques de modélisation par éléments finis afin de comprendre les mécanismes de la douleur cervicale. Certains ont soutenu que cet axe de recherche n'a pas abouti à des solutions tangibles au Canada. Cependant, d'autres ont utilisé la modélisation comme outil nécessaire à l'élaboration de solutions pour atténuer les douleurs au cou. Dans ce document, on soutient que les modèles conceptuels, physiques et empiriques peuvent être utilisés à diverses étapes pour aider à définir la portée du problème, de même que les tests de vérification et de validation appropriés, ainsi qu’à offrir une avenue pour l’élaboration, la conception et l'évaluation d’une solution. Cela a débouché sur une description du cadre HFM 252 de l'OTAN, ainsi que sur des équations mathématiques permettant d’évaluer la charge cervicale en fonction des caractéristiques de masse du système de casque. On en conclut que ces activités de modélisation sont essentielles à une recherche éthique et rentable sur les maux de cou.

Importance pour la défense et la sécurité

Ces travaux s’inscrivent dans le cadre du projet Efficacité humaine de la Force aérienne (03aa) du programme Souplesse aérienne de l’Aviation royale canadienne (ARC) et de RDDC, en vertu de l’élément de répartition du travail (ERT) « Solutions d’atténuation des problèmes de cou et de dos ». Ces travaux ont aussi des liens étroits avec ceux du Groupe de recherche sur les douleurs au cou chez les membres d'équipage de la Commission sur les facteurs humains et la médecine (FHM) FHM-252 de l’Organisation pour la recherche et la technologie (ORT) de l’Organisation du traité de l’Atlantique Nord (OTAN), de même qu'avec le projet concerté en médecine militaire sur l'atténuation de la douleur dans la moelle épinière chez le personnel navigant (Groupe technique 22) du Programme de coopération technique (TTCP).

L'un des résultats intermédiaires décrits dans le programme Souplesse aérienne consiste à « améliorer la santé et la sécurité du personnel opérationnel de l'ARC en étudiant les causes et les solutions relatives à des défis d’ordre aéromédical, comme l'hypoxie et les blessures découlant de la fatigue de la nuque ». (DND/CAF, 2013). Ce document traite de la façon d’utiliser les modèles et la modélisation pour l'élaboration et l'évaluation des solutions proposées pour les maux de cou qui affectent actuellement les équipages d'aéronefs. Il fournit le contexte approprié pour la modélisation afin que les développeurs et les évaluateurs puissent se concentrer sur l’élaboration de solutions, et non seulement sur la modélisation pour comprendre les mécanismes de la douleur cervicale.

Table of contents

List of figures

List of tables

Acknowledgements

The authors would like to acknowledge NATO Human Factors and Medicine 252 Research Task Group members who spent many hours developing the Aircrew Neck Pain Framework. We would also like to thank Dr. Marieke van den Oord and Mr. Yuval Steinman for their contributions to the Helmet Fit literature and solutions, which several countries have adopted including Canada. A special thank you to Mark Davies and GENTEX who provided the Ops-core skull mount and helmets for the Neck Supported Mass and Helmet Systems Assessment studies.

1 Introduction

Neck trouble[1] is a serious problem that impacts rotary wing and fast jet aircrew health and safety. In 2014, a survey was administered to Royal Canadian Air Force (RCAF) CH146 Griffon Helicopter aircrew, and 75% of pilots and Flight Engineers (FEs) reported various levels of neck pain (Chafé & Farrell, 2016). Of those who reported neck trouble, 23% were grounded, and 25% grounded themselves. At the same time, a Canadian Forces Health Services (CFHS) epidemiology study reported that the rate of incident and total primary care diagnoses of dorsalgia (common cervical pain) was 50.7 per 1000 person-years (Hawes, Whitehead, & Gray, 2014). These studies suggest that neck trouble impacts aircrew health and safety and may degrade task performance. Thus finding neck trouble mitigation solutions is imperative for mission effectiveness.

Defence Research and Development Canada (DRDC), Directorate of Technical Airworthiness and Engineering Support (DTAES), and Canadian Forces Environmental Medicine Establishment (CFEME) have been involved in neck trouble research for close to two decades (Xiao & Farrell, 2016). Some of the research used finite element and multi-body dynamics techniques to model neck structures (joints, muscles, ligaments, tendons, and disks) in order to understand neck pain mechanisms (Jahanshah et al., 2006; Jahanshah et al., 2005; Mustafy, El-Rich, Mesar, & Moglo, 2014). However, no operationally-practical solutions resulted from this modelling work. Decisions were made at the 2012 Neck Strain Workshop held at DRDC – Toronto Research Centre to redirect resources away from modelling studies, and initiate a project to find solutions that could be immediately implemented in operations. Even though mitigating solutions were found and delivered to the RCAF (Farrell et al., 2017), models were required during the solution proposal, development, and assessment phases of the project.

A model is a representation of a system often used to understand and predict its behaviour. The Technical Cooperation Program (TTCP) defines a model as a graphical, mathematical (symbolic), physical, or verbal representation or simplified version of a concept, phenomenon, relationship, structure, system, or an aspect of the real world (TTCP, unpublished). The DRDC Modelling and Simulation Community of Practice (M&S CoP) Terms of Reference define models as a representation of some aspect of the world expressed as variables and relationships between variables. Models may be stated in terms of prose (e.g., conceptual models), mathematically-based expressions, and empirical relationships (DRDC M&S CoP, unpublished).

This paper outlines three types of models that have been applied in neck trouble research. Chapter 2 introduces conceptual models, derived from logical reasoning, which identifies possible neck trouble factors. Chapter 3 highlights a simplified model, derived from physical laws of motion, which yield mathematical expressions for external neck loads. Chapter 4 describes empirical models, derived from survey and experimental data, which allows us to correlate reported neck pain to possible casual factors. Chapter 5 summarises how these types of models expedited the development of mitigation options and supported the conduct of ethical and cost effective neck-trouble research.

[1] For this paper, 'trouble' refers to one or more of the following: discomfort, fatigue, injury, acute pain, or chronic pain. Also, 'neck trouble' and 'neck pain' are used interchangeably in this report.

2 Conceptual models

Conceptual models are typically developed from logical reasoning and observation. Variables related to the concept are identified and defined, and relationships between these variables are described in prose (as opposed to mathematical symbology). In this section conceptual models are presented to understand 1) possible factors that cause neck trouble that yield a common taxonomy when proposing potential solutions, and 2) a conceptual relationship between external neck loads, muscle activity in response to these loads, and chronic neck pain.

2.1 Neck trouble possible factors

A conceptual model for neck trouble possible factors begins by identifying the variables or 'factors' that may contribute to aircrew neck-trouble.

Manufacturing assembly line work studies have shown that heavy helmet systems and repetitive tasks over long periods of time will cause injury and, if left unchecked, may result in chronic pain or behavioural adaptation to accommodate that pain (Guzman et al., 2009). We deduce that similar injuries may occur amongst aircrew when supporting heavy loads by their necks as they perform their flight tasks over several missions (Karakolis, Farrell, & Fusina, 2015). Also, it has been noted that poor musculoskeletal health and fitness might contribute to neck pain (Äng, 2007; Hébert, Roy, Burke, Côté, & Grodecki, 2012). Other factors may include age, gender, anthropometry, nutrition, sleep behaviour, non-flying activities (e.g., sport and office work), and pain history. Collectively, these aircrew **human factors** have been identified as possible contributors.

It is clear from various aircrew spinal studies that neck trouble depends on several factors involving the head and neck supported mass (*m*), Centre of Mass (*CoM*), and mass Moment of Inertia (*MoI*) properties of the helmet system (T. Bowden, 1987; Harrison, Neary, Albert, & Croll, 2012; Manoogian, Kennedy, & Duma, 2005; McLaughlin, 2013). A typical helmet system includes the helmet itself, Night Vision Goggles (NVG), Helmet Mounted Displays (HMDs), Counter Weight (CW), batteries and battery pack, communications equipment, and any other peripherals that are borne by the head and neck. The overall configuration of the helmet mounted visual systems tends to increase the forward neck flexion moment, increasing neck soft tissue stress. Thus, aircrew body-borne **equipment** (specifically neck-supported equipment) is a second factor that may contribute to aircrew neck-trouble.

Certain tasks performed by aircrew during flight result in awkward postures such as 'helo hunch' (T. J. Bowden, 1984; Maksel, 2014; Shanahan & Reading, 1984) where pilots tend to flex their upper back while their lower back is pressed flat against the seat—mainly due to seat, display, collective and cyclic designs. For example, Flight Engineers must sit and perform their tasks on unsupported 'rag-and-tube' seats that promote poor posture. Aircrew behaviours that may contribute to neck pain include performing tasks that involve holding relatively static postures for long periods of time or highly dynamic movements for short periods of time, extreme postures such as checking a slung load (Figure 1), and repetitive tasks such as visual scanning and Control Display Unit operations (Tack, Bray-Miners, Nakaza, Osborne, & Mangan, 2014). Performance of an individual task may result in acute pain that resolves itself without leading to chronic neck pain. However, performing these tasks while wearing head supported equipment over repeated long duration missions (exposure time) throughout a typical deployment, will likely increase the risk of neck pain. Thus, aircrew **behaviour** (tasks and postures) is a factor that potentially amplifies the impact of wearing heavy helmet systems.

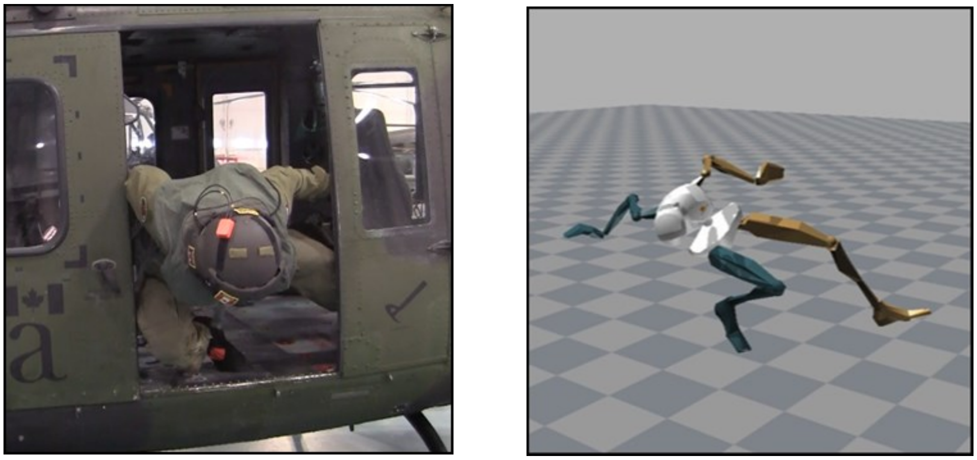

Figure 1: *'Checking Slung Load' extreme posture example with motion capture model equivalent (Tack et al., 2014) (Source: DRDC).*

As alluded to in the previous paragraph, poor seat design and suboptimal displays and controls may constitute a fourth factor that impacts neck trouble. In addition to these and other ergonomic workspace issues, aircrew are exposed to vibration and G force, which again amplify neck loads. This factor is called the aircraft **workspace** factor.

A fifth factor to be considered is the aircrew's **organisation** that makes decisions on mission objectives, intensity, duration, and frequency. Mission duration and the frequency of missions combine to determine exposure time, or the amount of time (organisation) aircrew (human factors) wear their helmet system (equipment) while performing tasks (behaviour) in the aircraft (workspace). Clearly relationships between the five factors begin to emerge.

A mission may require more training and operational sorties, fewer opportunities to rest, and more aggressive manoeuvres than usual. It is postulated that the cumulative neck load over several days, months, and years with insufficient recovery time between missions would eventually result in chronic neck trouble (Karakolis et al., 2015; Tack et al., 2014).

In summary, aircrew human factors (age, neck strength, gender), aircrew body-borne equipment (helmet system mass properties), behaviours (tasks and postures), workspace (vibrations, G forces and ergonomics), and organisation factors (mission type, duration, frequency) all combine to produce cumulative neck loading on neck structures over the course of a career. These five possible factors have been identified and discussed by Human Factors & Medicine (HFM) Research Task Group 252 on "Aircrew Neck Pain" (Farrell et al., draft). They are summarised in Table 1 and illustrated in Figure 2. Together these factors describe a framework (or a conceptual model), which in turn provides a common taxonomy and understanding of neck pain sources, and helps us target solutions that lessen the impact of these factors on aircrew neck pain.

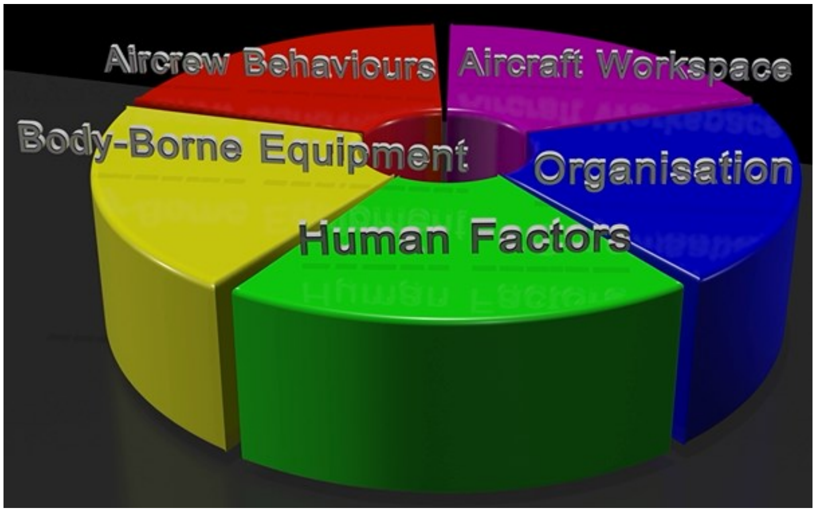

Figure 2: *Aircrew Neck Pain Possible Factors.*

Table 1: *HFM 252 Proposed framework for factors that impact aircrew neck trouble (Farrell et al., draft).*

Factor	Description
HUMAN FACTORS **Who Aircrew Are**	Human factors refer to the age, gender, anthropometry, strength, range of motion, muscle control, pain history and other personal physical characteristics that might be attributed to aircrew neck pain.
EQUIPMENT **What Aircrew Wear**	Equipment refers to the additional mass, inertia, and centre of mass due to head- and neck-borne equipment such as helmets, Night Vision Goggles, Helmet Mounted Displays, and Counter Weights.
BEHAVIOURS **What Aircrew Do**	Tasks and postures refer to not only the tasks that aircrew do during the course of a mission, but also the positions, postures, and postural sequences that they use while performing each task. Physical demands information (neck-supported mass properties and neck joint angles as a function of time) can be associated with each postural sequence, from which one may calculate the force and moment loads on neck joints.
WORKSPACE **Where Aircrew Work**	Seating, seat restraints, controls, instrumentation, rear cabin, and cockpit ergonomics may impact neck pain. Also, aircraft type may influence neck pain. That is, fast jets produce high-G, which amplifies any external neck loading, while helicopters produce vibration that adds a second order component to neck loading.
ORGANISATION **Why Aircrew Fly**	Organisation makes decisions with respect to the purpose and nature of the mission; e.g., search and rescue or tactical, day or night sorties, and long or short missions. It also makes decisions on organisational-level interventions; e.g., sortie scheduling, G-limiting standard operating procedures, Helmet Fit, and training.

There are two main concerns with this conceptual model. First, the model does not explicitly state the relationships between factors. For instance, external neck loads should increase as aircrew wear NVGs and perform extreme postural sequences in a vibrating and poorly ergonomically designed helicopter (Tack et al., 2014). In this statement, three factors—Equipment, Behaviours, and Workspace—combine to increase the risk of neck pain. Figure 2 does not adequately show the relationships between factors. Further concept development is required to tease out the relationships between factors.

The second issue is that this conceptual model has not been formally tested and evaluated. Model validation uses known data points within the modelling space and compares them to predicted data points generated by the model; the closer the predicted points are to the corresponding actual data points, the more confidence one has in the model to interpolate and extrapolate other data points within the modelling space. Testable postulates can be formulated from this conceptual model such as 'heavy helmet systems and poor posture lead to an increase in neck pain'. This postulate has been tested using survey data (Chafé & Farrell, 2016), modelling and simulation (Tack et al., 2014), and empirical data (Callaghan, 2014; Farrell et al., in press; Karakolis, McGuinness, & Farrell, draft). Other methods for evaluating concept models include testing the postulate against the scientific literature and subject matter expert opinion.

2.1.1 Proposed solutions that address causal factors

The primary use for this conceptual model in Table 1 and Figure 2 is to generate a useable and simple taxonomy that guides the development of neck trouble mitigating solutions. As solutions are being proposed, developed, tested, and implemented, the conceptual model can be further refined by asking questions such as 'are there any missing factors?' and 'are there any factors that need to be removed?'

Just as the possible aircrew neck pain causes are multi-factorial, so are neck pain mitigating solutions (Chafe, 2013; Fernie & Mayich, 2013; Fischer et al., 2013; NATO, 2013). Table 2 provides a list of proposed solutions currently being developed under the DRDC Air Human Effectiveness project; specifically the Neck- and Back-trouble Mitigating Solutions sub-project. The second column indicates the conceptual model factor(s) that the solution hopes to address. For example, the Canadian Surgeon General Exercise study assessed neck-specific exercises based on a study by Äng, Monnier, and Harms-Ringdahl (2009). These exercises address the human factors dimension, and in particular neck strengthening, flexibility, and stability, which are postulated to prevent or delay the onset of chronic neck pain (Hébert et al., 2012).

Table 2: *Proposed solutions and the factors they address.*

Proposed Solutions	**Factor(s) being Addressed**
Work-Rest Cycles	Organisation, Behaviour
Revised Workload Distribution	Behaviour
Helmet Fit	Organisation
Education and Exercise	Human factors
Neck Supported Mass Study	Equipment
Helmet System Support Devices	Equipment
Radar Altimeter Repeater Monitor	Behaviour, Workspace
"See through floor" capability	Workspace, Behaviour
Seat Ergonomics	Workspace
Passive and Active Vibration Mitigation Seats	Workspace

The set of proposed solutions covers the entire 'Factors' space where Workspace and Behaviour are noted four times, Organisation and Equipment two times, and Human factors once. However, this very simple analysis leads to more questions such as 'are the factors equally weighted with respect to their contribution to neck trouble?' 'Should solutions that address more than one factor be given more attention than others?' 'Are there any other solutions that might address human factors?'

Thus, the ‘neck trouble possible factors’ conceptual model is a useful framework that goes beyond understanding aircrew neck trouble and providing a taxonomy for exploring, designing, building, and assessing proposed solutions. It also can be used to pose key questions, postulates, and hypotheses that may lead to further studies and programs of work.

2.2 Chronic neck pain conceptual model

The relationship between the possible factors mentioned in the previous section and chronic neck pain is necessarily a conceptual model since it is difficult to establish causality relationships, e.g., between neck loads due to body-borne equipment and neck pain[2] (i.e., neck pain pathogenesis) for a number of reasons according to Moroney, Schultz, and Miller (1988). One reason is because chronic pain may develop as a result of repetitive or acute injuries. In both cases, long periods of time (months or years) are required to determine whether pain persists and becomes chronic. This makes it difficult to find a deterministic or empirical relationship between neck loads and chronic pain given that other life influences besides flying may trigger or exacerbate neck pain such as sleeping habits, office work, exercise, or an accident.

Despite the difficulty of finding a validated relationship between aircrew neck pain and its possible factors, one may still conceptualise a variety of logical postulates that may lead to a deeper understanding of the problem and guide in the development of solutions.

For instance, survey data yielded significant correlations between chronic neck pain and exposure time. A meta-analysis of Canadian, United Kingdom, and United States rotor-wing aircrew neck pain survey results “found that the dominant factors for predicting reported neck pain in aircrew are the total number of NVG hours and the average number of NVG hours per mission” (Fraser, Crowley, Shender, & Lee, 2015). That is, reports of neck pain occurred more frequently during the first 100 to 200 hours of NVG usage and after two hours of NVG usage per mission. Also, Karakolis et al. (2015) posit that chronic pain is a result of the overuse (exposure time) of neck structures where “cumulative loading of biological tissues exceeds the tissue’s ability to recover.”

Moreover, it has been shown that different types of loading on the spine cause different types of acute injury (McGill, 2007). Compression forces have been shown to be involved in stress fractures (Parkinson & Callaghan, 2007); shear forces in pars interarticularis damage (Howarth & Callaghan, 2013; Howarth, Karakolis, & Callaghan, 2015); and torque in herniated intervertebral discs (Tampier, Drake, Callaghan, & McGill, 2007). Also pain may be related to physical issues such as radiculopathy (nerve irritation or injury) or osteophytes (bone spurs), or related to metabolic response, i.e., fatigue and elevated pH (Pollak et al., 2014). Although these forms of structural damage can be observed with invasive and non-invasive techniques and likely produce acute pain, these injuries are not a predictor of chronic pain.

At least, one might postulate that cumulative neck loads associated with NVG usage may lead to higher muscle activation levels and injurious micro damage of neck structures[3] after a few hours of flying, and eventually chronic pain after 100 to 200 flight hours. This postulate may be expressed as a conceptual expression as follows:

---

[2] Conversely, there is a known causal relationship between neck loads and acute neck injuries (see **Table 3**).

[3] Micro tears of muscle fibre followed by healing are the normal process of muscle strengthening. Injurious micro damage refers to micro tears and stress fractures in soft and hard tissues that linger without the appropriate amount of time to heal before the next significant load exposure: i.e., the next mission.

$$\text{Chronic neck pain} = f\,(\text{exposure time, cumulative load, muscle activation, tissue damage}) \qquad (1)$$

Even though it may be possible to calculate/estimate/measure cumulative loads, muscle activation, tissue damage assessed using invasive and non-invasive techniques after each flight, and exposure time, neck pain is still subjective. That is, some aircrew reported anecdotally having a neck injury (i.e., damage to neck structures) without experiencing pain, while others continued to experience significant pain even after a known injury has healed[4].

Although, finding mathematical or empirical expressions may be formidable, if not unachievable for the above conceptual model in Equation 1 supported by the literature, we assert—based on this model—that proposed solutions that minimise exposure time and neck cumulative loads, and therefore muscle activation and tissue damage will reduce the risk of developing or aggravating chronic pain.

In summary, this chapter described two important conceptual models: a model that introduced possible factors of neck pain, and a model that relates (conceptually) chronic neck pain to exposure time, cumulative load, muscle activation, and tissue damage. These conceptual models are useful for framing and understanding the problem, providing a common taxonomy for discussing the problem as well as categorising solutions, and generating testable hypotheses that have been explored empirically. Some shortfalls of these models are that the relationships between the model variables are not fully developed, and these conceptual models still require validating; experimentally, with literature sources, or with subject matter expert opinion. Nevertheless, conceptual models have been used as a starting point for exploring aircrew neck-trouble solutions.

[4] Personal Communication with Griffon Helicopter aircrew at 404 Squadron, Valcartier, Quebec. September, 2013.

3 Physics-based models

The literature and biomechanical software such as Visual 3D contain mathematical expressions for neck joint kinematics and dynamics but for no head supported mass; e.g., (Kingma, de Looze, Toussaint, Klijnsma, & Bruijnen, 1996; Moglo, Mesfar, & Mustafy, 2012; Mustafy et al., 2014; Vette, Yoshida, Thrasher, Masani, & Popovic, 2012). Other papers logically deduce and measure how muscle activity and neck loads respond to head supported mass; e.g., (Burnett, Green, Netto, & Rodrigues, 2007; Callaghan, 2014; Dibblee et al., 2015; Fischer, Stevenson, Reid, & Hetzer, 2014; Forde et al., 2011; Harrison, Coffey, Albert, & Fischer, 2015; Moroney et al., 1988). Rarely are mathematical expressions used to support these deductions, to the author's knowledge. This chapter derives a simplified mathematical model of a generic neck supported helmet system to support the logical deduction that helmet system mass properties impact neck loads.

In this chapter, "physics-based" means that the relationship between helmet system mass properties and neck loading may be expressed in mathematical terms by applying Newton's second law of motion with simplifying assumptions of varying degrees. These equations provide insight into how neck loads change as a function of neck-borne equipment, which may lead to certain solutions that reduce cumulative neck loading and potentially minimise the risk of developing or aggravating chronic pain.

The dependent variables for this physics-based model are the neck loads (i.e., compression force, shear force, and torque) at a given neck joint. The independent variables include helmet fit frictional forces, and head and helmet system mass properties: *m*, *CoM*, and *MoI*. In this case, the generic helmet system includes a helmet and chin straps, Night Vision Goggles (NVG), and Counter Weight and battery pack combination (CW) as shown in Figure 3. Other independent variables, such as communication equipment and atmospheric conditions, were omitted from this modelling exercise for simplicity, but may be added subsequently to increase model fidelity as needed.

3.1 Head and helmet system simplified physics-based model

3.1.1 Simplified single neck joint load equations

Neck load expressions have been derived from mechanics and biomechanics considerations (Vette et al., 2012). Here we will derive a model for a single neck joint and structure supporting a generic head/helmet system as shown in Figure 3: that is, two solid bodies pivoting around a single joint. The head ($\boldsymbol{F}_h$), Helmet ($\boldsymbol{F}_H$), NVG ($\boldsymbol{F}_{NVG}$), and CW ($\boldsymbol{F}_{CW}$) forces act on the head/ helmet system body through their *CoM*. Also, the friction force due to the helmet fit ($\boldsymbol{F}_{HF}$) acts tangentially at the head/Helmet interface, chin strap forces ($\boldsymbol{F}_p$ and $\boldsymbol{F}_q$) contribute to $\boldsymbol{F}_{HF}$ without affecting joint loads, and muscle forces ($\boldsymbol{F}_M$) act directly on the joint and create a restoring muscle torque ($\boldsymbol{T}_M$).

In this simplified model, we will consider the forces acting on the seventh cervical—first thoracic (C7) neck joint for ease of comparison with the Integrated Mission function task analysis / Physical demands analysis Model (IMPM) study (Tack et al., 2014). The assumptions are:

1. All objects are solid bodies, and neither mass nor inertia change over time;
2. C7 joint is in static equilibrium, thus resultant forces and torques acting on C7 are zero;

3. C7 joint is a point mass; and
4. All objects are symmetrical.

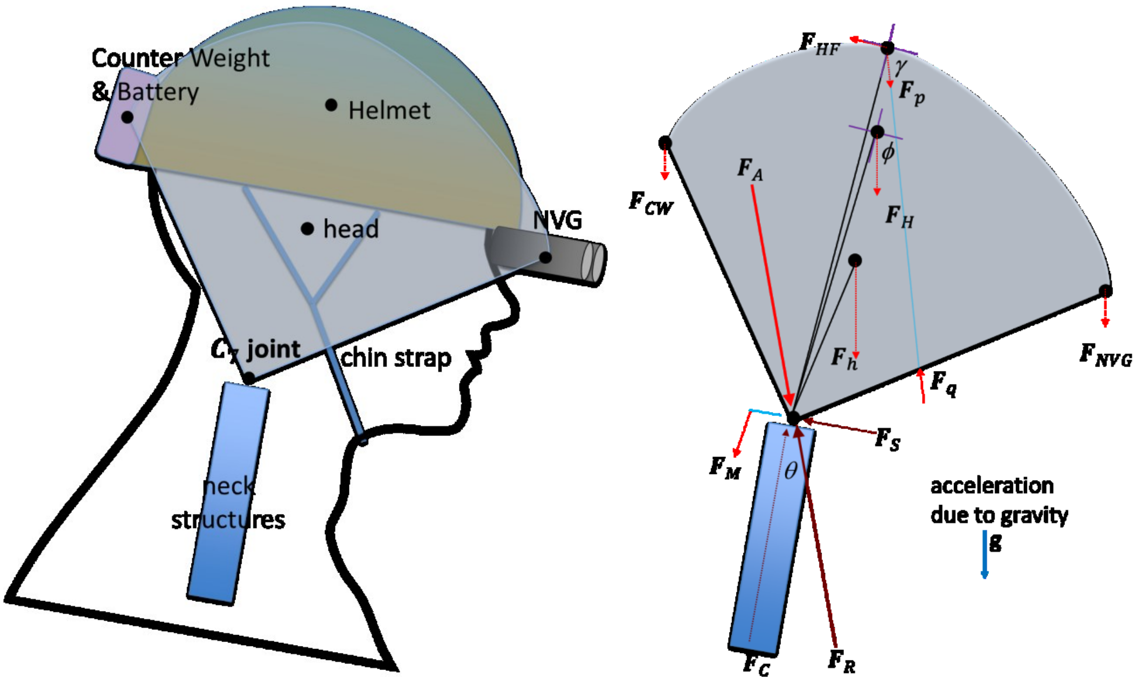

__Figure 3:__ Simplified model of generic head/Helmet system and neck structure pivoting around a C7 joint. The free-body diagram shows action and reaction forces acting at the joint.

Newton's 2nd Law of Motion and Conservation of Linear and Angular Momentum expressions states that the sum of forces is equal to the change of momentum and the sum of torques is equal to the change in angular momentum (Meriam, 1980):

$$\sum \boldsymbol{F} = \frac{d(m\boldsymbol{v})}{dt} = m\boldsymbol{a} + \boldsymbol{v}\frac{dm}{dt} \tag{2}$$

and

$$\sum \boldsymbol{T} = \frac{d([\mathbf{I}]\boldsymbol{\omega})}{dt} = [\mathbf{I}]\boldsymbol{\alpha} + \boldsymbol{\omega}\frac{d[\mathbf{I}]}{dt} \tag{3}$$

where $\sum \boldsymbol{F}$ is the sum of forces acting on an object, *m* is the mass of the object, ***v*** is the linear velocity vector and ***a*** is the linear acceleration vector due to linear velocity changes (including G = a_z/g) and vibration of the objects. $\sum \boldsymbol{T}$ is the sum of torques acting on the object, *[I]* is the inertia tensor of the object, **ω** is the angular velocity, and **α** is the angular acceleration vector of the object. (***Bold*** and *Italic* fonts indicate **vector** and *scalar* quantities, respectively.) Applying assumptions 1 and 2 to Equations 2 and 3 yields:

$$\sum \boldsymbol{F} = 0 \tag{4}$$

and

$$\sum \boldsymbol{T} = 0 \tag{5}$$

3.1.1.1 Compression and shear forces acting on single neck joint

The vector sum of forces acting on the head/helmet system body equals to an Action Force ($\boldsymbol{F}_A$). $\boldsymbol{F}_A$ and its Reaction Force ($\boldsymbol{F}_R$) act on the C7 joint. Substituting $\boldsymbol{F}_A$ and $\boldsymbol{F}_R$ into Equation 4 yields:

$$\sum \boldsymbol{F} = 0 = \boldsymbol{F}_A - \boldsymbol{F}_R \tag{6}$$

$\boldsymbol{F}_A$ is the vector sum of $\boldsymbol{F}_M$, $\boldsymbol{F}_{HF}$ as well as the forces due to gravity ($\boldsymbol{g}$) produced by the mass of each neck supported object: head (m_h), helmet (m_H), NVG (m_{NVG}), and CW (m_{CW}). $\boldsymbol{F}_{HF}$ may be expressed as follows:

$$\boldsymbol{F}_A = \boldsymbol{F}_h + \boldsymbol{F}_H + \boldsymbol{F}_{NVG} + \boldsymbol{F}_{CW} + \boldsymbol{F}_{HF} + \boldsymbol{F}_M$$

$$\boldsymbol{F}_A = m_h\boldsymbol{g} + m_H\boldsymbol{g} + m_{NVG}\boldsymbol{g} + m_{CW}\boldsymbol{g} + \boldsymbol{F}_{HF} + \boldsymbol{F}_M$$

$$\boldsymbol{F}_A = (m_h + m_H + m_{NVG} + m_{CW})\boldsymbol{g} + \boldsymbol{F}_{HF} + \boldsymbol{F}_P + \boldsymbol{F}_M$$

$$\boldsymbol{F}_A = m_T\boldsymbol{g} + \boldsymbol{F}_{HF} + \boldsymbol{F}_M \tag{7}$$

where m_T is the total neck supported mass. Note that the chin strap forces are equal and opposite to each other ($\boldsymbol{F}_p = -\boldsymbol{F}_q$) acting along the same line of action, and therefore do not contribute to either $\boldsymbol{F}_A$ or $\boldsymbol{T}_M$. However, $\boldsymbol{F}_p$ generates an additional normal force that contributes to the tangential frictional force between the helmet system and the head, $\boldsymbol{F}_{HF}$. $\boldsymbol{F}_{HF}$ prevents the helmet from slipping relative to the head. The magnitude of $\boldsymbol{F}_{HF}$ is the coefficient of friction (μ) multiplied by the normal component of both the sum of helmet system forces as well as the force that the helmet exerts on the head ($\boldsymbol{F}_p$) and the head exerts on the helmet ($\boldsymbol{F}_q$) due to the chin strap ($2F_P\cos\gamma$) plus force components acting in the tangential direction. Thus, $\boldsymbol{F}_{HF}$ is as follows:

$$\boldsymbol{F}_{HF} = -\mu[(m_H + m_{NVG} + m_{CW})g\sin\phi + 2F_P\sin\gamma]\boldsymbol{t} + [(m_H + m_{NVG} + m_{CW})g\cos\phi + 2F_P\cos\gamma]\boldsymbol{t}$$

$$\boldsymbol{F}_{HF} = -\mu\left[m_{HS}g\left(\sin\phi - \tfrac{1}{\mu}\cos\phi\right) + 2F_P\left(\sin\gamma - \tfrac{1}{\mu}\cos\gamma\right)\right]\boldsymbol{t} \tag{8}$$

where m_{HS} is the total helmet system mass, g is the magnitude of $\boldsymbol{g}$, and $\boldsymbol{t}$ is a unit vector in the tangential direction. Substituting Equation 8 into Equation 7 yields:

$$\boldsymbol{F}_A = m_T\boldsymbol{g} - \left[m_{HS}g\left(\sin\phi - \tfrac{1}{\mu}\cos\phi\right) + 2F_P\left(\sin\gamma - \tfrac{1}{\mu}\cos\gamma\right)\right]\boldsymbol{t} + \boldsymbol{F}_M \tag{9}$$

$\boldsymbol{F}_R$ may be resolved into Compression (F_C) and Shear (F_S) force components acting on the C7 joint supported by neck structures as follows:

$$F_C = \|\boldsymbol{F}_R\|\cos\theta \tag{10}$$

$$F_S = \|\boldsymbol{F}_R\|\sin\theta \tag{11}$$

where θ is the angle between $\boldsymbol{F}_R$ and the C7 joint longitudinal axis. From Equation 6, $\boldsymbol{F}_R = \boldsymbol{F}_A$. Substituting Equation 9 into Equations 10 and 11 yields $\boldsymbol{F}_C$ and $\boldsymbol{F}_S$ in terms of the total neck supported mass, acceleration due to gravity, helmet/head friction coefficient, helmet mass, and muscle forces as follows:

$$F_C = \left\| m_T \boldsymbol{g} - \left[m_{HS} g \left(\sin\phi - \frac{1}{\mu}\cos\phi \right) + 2F_P \left(\sin\gamma - \frac{1}{\mu}\cos\gamma \right) \right] \boldsymbol{t} + \boldsymbol{F}_M \right\| \cos\theta \quad (12)$$

$$F_S = \left\| m_T \boldsymbol{g} - \left[m_{HS} g \left(\sin\phi - \frac{1}{\mu}\cos\phi \right) + 2F_P \left(\sin\gamma - \frac{1}{\mu}\cos\gamma \right) \right] \boldsymbol{t} + \boldsymbol{F}_M \right\| \sin\theta \quad (13)$$

Relaxing the assumption that C7 is in static equilibrium, Equation 6 may be expressed as follows:

$$\boldsymbol{F}_A - \boldsymbol{F}_R = m_{C7} \boldsymbol{a} \quad (14)$$

where m_{C7} is C7 joint mass. Thus, the compression and shear force expressions become:

$$F_C = \left\| (m_{C7} + m_T) \boldsymbol{a} - \left[m_{HS} a \left(\sin\phi - \frac{1}{\mu}\cos\phi \right) + 2F_P \left(\sin\gamma - \frac{1}{\mu}\cos\gamma \right) \right] \boldsymbol{t} + \boldsymbol{F}_M \right\| \cos\theta \quad (15)$$

$$F_S = \left\| (m_{C7} + m_T) \boldsymbol{a} - \left[m_{HS} a \left(\sin\phi - \frac{1}{\mu}\cos\phi \right) + 2F_P \left(\sin\gamma - \frac{1}{\mu}\cos\gamma \right) \right] \boldsymbol{t} + \boldsymbol{F}_M \right\| \sin\theta \quad (16)$$

where $\boldsymbol{a}$ is the linear acceleration vector due to linear velocity changes and is the same for all objects in the model (a is the magnitude of $\boldsymbol{a}$). The third assumption—C7 joint is a point mass—means that if the volume is infinitesimally small then the mass must be just as small to maintain the same joint density. Thus, Equations 15 and 16 simplify even further to:

$$F_C = \left\| m_T \boldsymbol{a} - \left[m_{HS} a \left(\sin\phi - \frac{1}{\mu}\cos\phi \right) + 2F_P \left(\sin\gamma - \frac{1}{\mu}\cos\gamma \right) \right] \boldsymbol{t} + \boldsymbol{F}_M \right\| \cos\theta \quad (17)$$

$$F_S = \left\| m_T \boldsymbol{a} - \left[m_{HS} a \left(\sin\phi - \frac{1}{\mu}\cos\phi \right) + 2F_P \left(\sin\gamma - \frac{1}{\mu}\cos\gamma \right) \right] \boldsymbol{t} + \boldsymbol{F}_M \right\| \sin\theta \quad (18)$$

Other helmet system components may be added to, or subtracted from, the equations above. For example, $m_T = m_h + m_H$ and $m_{HS} = m_H$ (i.e., no NVG or CW) may be substituted into Equations 17 and 18 to yield the compression and shear forces for a day flight helmet system configuration.

3.1.1.2 Torque acting on single neck joint

Neck muscles generate the necessary torque to achieve static equilibrium for all head positions (Moroney et al., 1988). This muscle torque ($\boldsymbol{T}_M$) is equal and opposite to the sum of torques generated by forces shown in Figure 4. The torque generated by the ith object's force may be expressed as follows (Meriam, 1980):

$$\boldsymbol{T}_i = [\boldsymbol{I}]_i \boldsymbol{\alpha} + \boldsymbol{F}_i \times \boldsymbol{d}_i$$

$$\boldsymbol{T}_i = [\boldsymbol{I}]_i \boldsymbol{\alpha} + m_i g \, d_i \sin\theta_i \, \boldsymbol{i_i} \quad (19)$$

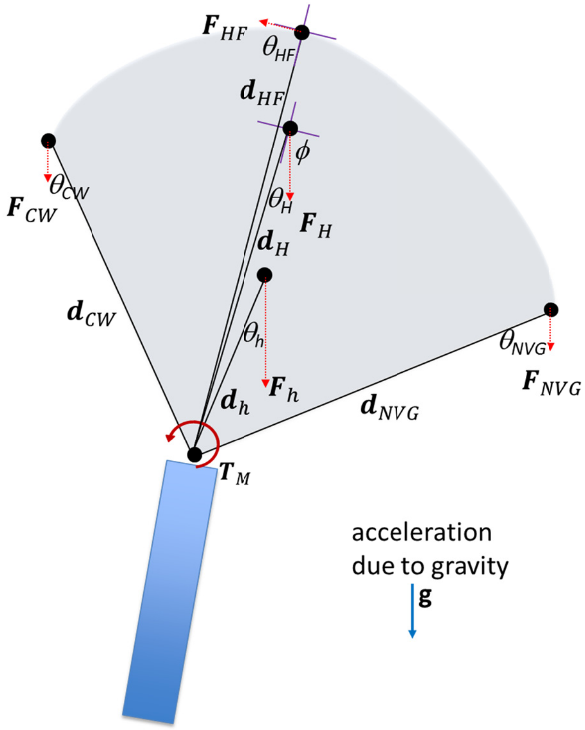

Figure 4: *Torques (Moments) acting on C7Joint. Torques due to* $\boldsymbol{F}_{CW}$ *and* $\boldsymbol{F}_{HF}$ *act in the same direction as* $\boldsymbol{T}_M$ *for this posture while* $\boldsymbol{F}_h$*,* $\boldsymbol{F}_H$ *and* $\boldsymbol{F}_{NVG}$ *provide a counter-torque.*

where $[\boldsymbol{I}]_i$ is the i$^{\text{th}}$ object's inertia tensor, $\boldsymbol{\alpha}$ is its angular acceleration (assumed to be zero for static equilibrium assumption), $\boldsymbol{F}_i$ is the object's force due to gravity ($m_i\boldsymbol{g}$), $\boldsymbol{d}_i$ is the displacement between C7 and the object's CoM. g and d_i are the magnitudes of $\boldsymbol{g}$ and $\boldsymbol{d}_i$, respectively. $\boldsymbol{i}_i$ is a unit vector perpendicular to both $\boldsymbol{F}_i$ and $\boldsymbol{d}_i$ as a result of the cross product that obeys the 'left-hand rule'($\boldsymbol{F}_i$ onto $\boldsymbol{d}_i$). The individual θ_i's are a function of the head position relative to the thoracic segment; that is, the C7 joint angular position. Thus, substituting $\boldsymbol{T}_i$ into Equation 5 and assuming $\boldsymbol{\alpha} = \boldsymbol{0}$ yields (Note well that the resultant torque due to $\boldsymbol{F}_P$ and $\boldsymbol{F}_q$ is zero since they pass through the same line of action and effectively cancel out each other):

$$\sum \boldsymbol{T} = \boldsymbol{0} = -\boldsymbol{T}_M - \mu m_{HS} g \left[\sin\phi - \frac{1}{\mu}\cos\phi\right] d_{HF} \sin\theta_{HF}\, \boldsymbol{i}_{HF} - m_{CW} g d_{CW} \sin\theta_{CW}\, \boldsymbol{i}_{CW}$$

$$+ m_h g d_h \sin\theta_h\, \boldsymbol{i}_h + m_H g d_H \sin\theta_H\, \boldsymbol{i}_H + m_{NVG} g d_{NVG} \sin\theta_{NVG}\, \boldsymbol{i}_{NVG} \quad (20)$$

Consider a 2-dimensional case where displacement and acceleration are projected onto the y-z plane, and $\boldsymbol{i}$ is a unit vector in the x direction perpendicular to the y-z plane. The x-axis represents the flexion/extension axis of rotation. Equation 20 simplifies to:

$$\boldsymbol{T}_{M_x} = \Big(m_h g d_{h_{yz}} \sin\theta_{h_x} + m_H g d_{H_{yz}} \sin\theta_{H_x} + m_{NVG} g d_{NVG_{yz}} \sin\theta_{NVG_x}$$

$$- \mu m_{HS} g \left[\sin\phi - \frac{1}{\mu}\cos\phi\right] d_{HF_{yz}} \sin\theta_{HF_x} - m_{CW} g d_{CW_{yz}} \sin\theta_{CW_x} \Big)\, \boldsymbol{i} \quad (21)$$

Similar expressions are found for y and z components of $\boldsymbol{T}_M$. Thus, when the C7 joint is in static equilibrium, the muscle torque depends on the mass, acceleration due to gravity, distances between each object and C7 joint, angle between gravity and displacement vectors for each object, as well as the helmet fit coefficient of friction.

Allowing the C7 joint to move (i.e., $\boldsymbol{\alpha} \neq \mathbf{0}$ and $\mathbf{a} \neq \mathbf{0}$), the inertia term in Equation 19 becomes part of the torque expression. Also, the fourth assumption states that objects are symmetrical (i.e., off-diagonal inertia tensor terms are zero). Recall that C7 joint is assumed to be a point mass; that is, $[\boldsymbol{I}_{C7}] = [\mathbf{0}]$. Given these assumptions, consider a pure flexion/extension angular acceleration around the x-axis. Note that $\boldsymbol{F}_{HF}$ does not have a volume associated with it, and therefore $[\boldsymbol{I}_{HF}] = [\mathbf{0}]$. Given that $I_{xx} = I_{h_{xx}} + I_{H_{xx}} + I_{NVG_{xx}} + I_{CW_{xx}}$ is the sum of each object's inertia, an additional inertia term is added to Equation 21as follows:

$$\boldsymbol{T}_{M_x} = \Big(I_{xx}\alpha_x + m_h a_{yz} d_{h_{yz}} \sin\theta_{h_x} + m_H a_{yz} d_{H_{yz}} \sin\theta_{H_x} + m_{NVG} a_{yz} d_{NVG_{yz}} \sin\theta_{NVG_x}$$
$$-\mu m_{HS} a_{yz} d_{HF_{yz}} \left[\sin\phi - \tfrac{1}{\mu}\cos\phi\right] \sin\theta_{HF_x} - m_{CW} a_{yz} d_{CW_{yz}} \sin\theta_{CW_x}\Big)\,\boldsymbol{i} \quad (22)$$

Similar expressions are found for y and z components of $\boldsymbol{T}_M$. Thus, the muscle torque vector may be expressed as follows:

$$\boldsymbol{T}_M = \left(\begin{bmatrix} I_{xx} & 0 & 0 \\ 0 & I_{yy} & 0 \\ 0 & 0 & I_{zz} \end{bmatrix}\begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} + \begin{bmatrix} mad_{h_{yz}} & mad_{H_{yz}} & mad_{NVG_{yz}} & -\mu mad_{HS_{yz}} & -mad_{CW_{yz}} \\ mad_{h_{xz}} & mad_{H_{xz}} & mad_{NVG_{yz}} & -\mu mad_{HS_{xz}} & -mad_{CW_{xz}} \\ mad_{h_{xy}} & mad_{H_{xy}} & mad_{NVG_{xy}} & -\mu mad_{HS_{xy}} & -mad_{CW_{xy}} \end{bmatrix}\begin{bmatrix} s\,\theta_{h_z} & s\,\theta_{h_y} & s\,\theta_{h_x} \\ s\,\theta_{H_z} & s\,\theta_{H_y} & s\,\theta_{H_x} \\ s\,\theta_{NVG_z} & s\,\theta_{NVG_y} & s\,\theta_{NVG_x} \\ c\phi_r\, s\theta_{HS_z} & c\phi_r\, s\theta_{HS_y} & c\phi_r\, s\theta_{HS_x} \\ s\,\theta_{CW_z} & s\,\theta_{CW_y} & s\,\theta_{CW_x} \end{bmatrix}\right)\begin{bmatrix} \boldsymbol{i} \\ \boldsymbol{j} \\ \boldsymbol{k} \end{bmatrix} \quad (23)$$

Where $mad_{n_{pl}}$ is a short form for $m_o a_{pl} d_{o_{pl}}$, *o* refers to the object, and *pl* refers to the plane of rotation. Also, $s\,\theta_{o_r}$ is a short form for $\sin\theta_{o_r}$ where *r* refers to the axis of rotation. Finally, $c\phi$ is a short form for $[\sin\phi_r - (1/\mu)\cos\phi_r]$. Thus, the muscle force is related to the moment of inertia, angular acceleration, object mass, linear acceleration, and distance from the object to C7 joint, the joint angle[5], as well as the coefficient of friction.

3.1.2 Solution implications from neck load equations

The shear and compression forces (Equations 17 and 18) describe a linear relationship between the total neck-supported mass (m_T) and the linear acceleration vector ($\boldsymbol{a}$). That is, reducing m_T or $\boldsymbol{a}$ will reduce F_C and F_S. Thus, the equations support the logical deduction that less mass, vibration, and G (= $\boldsymbol{a}/g$) reduce neck forces, which in turn suggests lighter helmet system and vibration mitigation solutions. The

[5] Since the head/helmet system is modelled as a single body, all angles are fixed relative to each other, and together they change relative to the joint angle.

equations also indicate that maximising the coefficient of friction contributes to a reduction of compression and shear loads. Likewise, $\boldsymbol{F}_p$ contributes to compression and shear forces indirectly through the frictional force by increasing the normal force. Note that γ is relatively fixed near to 90^o and, therefore, its tangential component is small.

As mentioned, the C7 joint angle (not shown) is the head/ helmet system segment angle relative to the neck structure segment. The C7 joint angle is between the joint longitudinal axis ($\boldsymbol{F}_C$ direction) and the head displacement vector, $\boldsymbol{d}_h$. In Figures 3 and 4, the joint angle is about 10^o. As the joint angle approaches zero, the head is in an upright and eyes forward position, shear forces approach zero, and the compression force equal to the reaction force.

This is a biomechanically advantageous position since neck structures can support 3 to 6 times more force in the compression direction than in the shear direction before reaching a structural limit (see Table 3). Thus, solutions that promote biomechanically advantageous postures will minimise shear force loading.

Table 3: *Neck Structural limits (Blackmore, Goswami, & Chancey, 2012).*

Extension Bending Moment at occipital condyle	47.3 Nm Pain threshold 56.7 Nm Significant structural damage
Flexion Bending Moment at occipital condyle	59.4 Nm Pain threshold 189 Nm Significant structural damage
Compression (Extension)	1720 ± 1230 N Bilateral facet dislocations
Compression (Flexion)	2000 ± 1230 N Bilateral facet dislocations
Pure Compression	4810 ± 1290 N
Tension Cervical Spine	1135 N (some evidence from past studies)
Tension Intervertebral Discs	581 ± 220 N at failure
Axial Torsion	Lower bound between 13.6 ± 4.5 Nm and 17.2 ± 5.1 Nm
Horizontal Shear	824 N Transverse ligament rupture with anterior shear of the atlas
Horizontal Shear	1510 ± 420 N Odontoid fractures

There are a number of solution implications from the muscle torque Equation 23. Starting from the left of the equation, muscle torque is minimised by reducing the sum of inertias along each rotation axis (I_{xx}, I_{yy}, and I_{zz}). Minimising an object's inertia involves reducing the volume or profile of the helmet, NVGs, or CW and battery pack. For example, a low-profile helmet system might involve denser CWs that provide the same counterbalance but its volume would be smaller. In fact, the CW and battery pack could be integrated into the existing helmet shell thus reducing the profile even further[6]. Low-profile NVGs would also lead to reduced muscle torque.

[6] An NSERC DND Grant has been awarded in July 2017 to investigate Next Generation Helmet Systems with counterbalance integrated into the helmet shell.

Muscle torque is minimised when angular accelerations ($\boldsymbol{\alpha}$) are small, and thus slow rotational neck movements are preferred. For instance, aircrew might be encouraged to bend at the waist rather than the neck and rotate the head and shoulders rather than the neck alone whenever possible.

Linear acceleration appears again in Equation 23. Note that when the helicopter is in a hover or cruising (i.e., small linear velocity changes), the acceleration due to gravity ($\mathbf{a} \approx \mathbf{g}$) dominates, and muscles still work to maintain the head and helmet system in a static posture. Nevertheless, any aircraft workspace solution that minimises linear velocity changes, including those associated with vibration[7] and G > 1 would be beneficial; recognising that these solutions would likely require costly aircraft modifications.

Equation 23 shows that muscle torque varies positively and linearly with head, helmet, and NVG mass. Thus, minimising m_H and m_{NVG} reduces $\|\boldsymbol{T}_M\|$. Any solution that involves lighter helmets and NVGs will reduce neck loads, all else being equal.

The helmet fit term associated with m_{HS} is negative. That is, if the helmet system wants to slip in the flexion/extension, lateral bending, or axial rotation directions, frictional forces will resist these movements for an appropriately large value of μ. However, as μ approaches zero (i.e., a loose helmet fit), frictional forces become small and ineffective, and neck muscle torque must compensate to prevent the helmet from slipping. Thus, a helmet liner with a relatively high friction coefficient would be beneficial in lowering neck torques and avoiding slippage.

Dead weight on the head is not enough to prevent the most helmets from slipping. Thus, chin straps are used to increase the downward normal force of the helmet with respect to the head ($2F_P\cos\gamma$) and therefore increase $\boldsymbol{F}_{HF}$. However, if the straps are too tight (high F_p), this may lead to hotspots and pressure points. And so, a proper helmet fit provides an optimal strap tension that prevents slippage while avoiding hotspots and pressure points.

The last term in Equation 23 is the sine matrix. As described above, this matrix varies as a function of joint angle. We shall see in Section 3.1.2.1 that all sine terms are positive in a neutral posture only, but as the head rotates around the positive x-axis (flexion), $\sin\theta_{CW_x}$ passes through zero and becomes negative and the neck torque must work even harder to counter the CW.

3.1.2.1 Counter weight effects

Although various studies have investigated the use of counterweights (Chafé & Farrell, 2016; Tack et al., 2014; van den Oord, 2012) the debate regarding them continues. Some aircrew will not fly with them, and others will not fly without them. Equations 17 and 18 show clearly that as m_{CW} increases, F_C and F_S increase linearly. However, Equation 23 shows that CW will reduce $\|\boldsymbol{T}_M\|$ as long as:

- $s\,\theta_{CW_z}$ is positive, then the torque is reduced around the flexion/extension axis.
- $-mad_{CW_{xz}}$ remains negative: i.e., NVGs are centred and above the y-axis and CWs are centred and below the y-axis, reducing torque around the lateral bending axis.
- CW mass in $-mad_{CW_{yz}}$ is reduced, reducing torque around the axial rotation axis.

[7] Projects are underway with National Research Council of Canada (NRCC) investigating passive and active seat vibration mitigation technologies (Chen, Wickramasinghe, & Zimcik, 2011), including a joint collaboration with the United States Navy.

For instance, consider a pure flexion position for joint angles 0°, 30°, and 45° as shown in Figure 5. Appropriately scaled values (from the figures) are provided in Table 5 and substituted into Equation 21 to illustrate the detrimental effect of CW as it passes through the vertical axis.

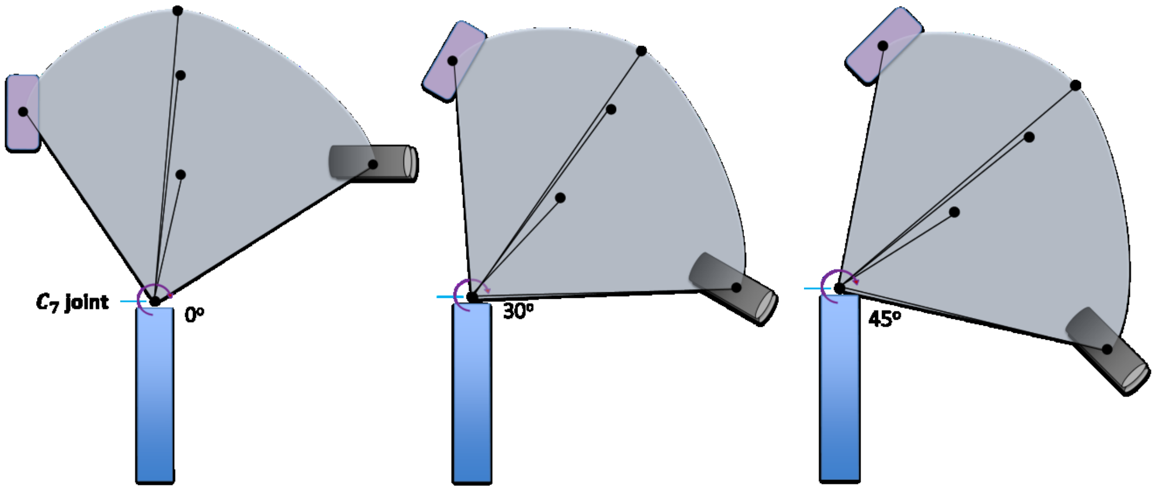

Figure 5: *Head/Helmet System body position relative to neck structure body for. flexion rotation only, and for 0°, 30°, and 45° C7 joint angles.*

Table 4: *Muscle Torque for 0°, 30°, and 45° C7 joint angles with and without CW.*

	NVG	head	helmet	HF	CW	with CW	without CW
mass (kg)	0.2	6.0	2.0	2.8	0.6	8.8	8.2
distance (cm)	9.4	4.6	8.3	10.7	8.4		
acceleration (m/s^2)	9.8	9.8	9.8	9.8	9.8		
μ				0.2			
object angle when C7 joint angle is **0°**	58°	13°	6°	5°	34°	$\|\|T_M\|\|$ (Equation 21)	
Torque (Nm)	0.1	0.6	0.2	-0.2	-0.3	0.4	0.7
object angle when C7 joint angle is **30°**	88°	43°	36°	35°	4°		
Torque (Nm)	0.1	1.8	1.0	1.3	0.0	4.2	3.9
object angle when C7 joint angle is **45°**	103°	58°	51°	50	-11°		
Torque (Nm)	0.1	2.3	1.3	1.9	0.1	5.7	5.2

With CW, the muscle torque is 0.4, 4.2, and 5.7 Nm when the C7 joint angle is 0°, 30°, and 45°, respectively. The 30° and 45° positions require 10 and 14 times more muscle torque, respectively, than when the joint angle is 0° (e.g., flying pilot tends to spend most of their time scanning in a relatively neutral position). This derivation clearly shows that neck posture contributes to neck loading.

Note that for a 30° joint angle, CW restoring torque is zero, and for 45° CW detrimentally adds to the muscle torque (e.g., non-flying pilot flexes and rotates their neck downward to operate the Control Display Unit over the course of a mission). Also note that the torque due to $\boldsymbol{F}_{HF}$ is only beneficial in a heads up neutral position. As the head/helmet system moves into a flexion posture, $m_{HS}g\cos\phi$ dominates and negates any frictional forces, and the muscles must work even harder to maintain a stable position. This situation becomes even worse when $\phi = 0$ and neck muscles must work at their maximum to counter the force due to the entire head borne mass ($m_{HS}g$) (e.g., Flight Engineers experience the highest loads when they lie in a prone position out the door looking down to check a slung load).

Without CW, the muscle torque is 0.7, 3.9, and 5.2 Nm when the C7 joint angle is 0°, 30°, and 45°, respectively. The 30° and 45° positions require only 5 and 7 times more muscle torque, respectively, than when the joint angle is 0°. Note that 'without CW' muscle torque 43% higher in a neutral posture compared to the 'with CW' condition. However, the overall torque is relatively small in both cases. In a 30° and 45° flexion posture, 'without CW' provides 7% and 9% reduction in muscle torque, respectively, compared to 'with CW'. Recall that having no CW also reduces compression and shear forces for all postures.

Thus, counterbalance is effective only when the head is in relatively neutral and extension[8] positions. Any solution that provides counterbalance without using counter weights (m_{CW} = 0), such as spring-loaded mechanisms (Fischer et al., 2014) or chin rests (Tack, Bray-Miners, Nakaza, & Osborne, 2016), would reduce muscle torque. In general, CW is a sub-optimal solution for reducing neck loads. Care must be taken in how they are deployed. For example, flying pilots might benefit slightly from counter weights, but CW would provide less benefit for non-flying pilots and FEs.

3.1.2.2 Centre of mass and counter weight

So far, this model has provided insight into C7 joint compression and shear forces and torque in terms of mass and *MoI*. This subsection investigates the impact of *CoM* on joint neck loads. *CoM* is calculated as follows:

$$CoM = \frac{m_h\boldsymbol{d}_h + m_H\boldsymbol{d}_H + m_{NVG}\boldsymbol{d}_{NVG} + m_{CW}\boldsymbol{d}_{CW}}{m_h + m_H + m_{NVG} + m_{CW}} \tag{24}$$

Recall that at the C7 joint, positive x-axis is transverse axis pointing right, the positive y-axis is the anterior/posterior axis pointing front, and the positive z-axis is the longitudinal axis pointing upwards. Thus, the *CoM* coordinates are as follows:

$$CoM_x = \frac{m_h d_{h_x} + m_H d_{H_x} + m_{NVG} d_{NVG_x} + m_{CW} d_{CW_x}}{m_h + m_H + m_{NVG} + m_{CW}} = 0 \tag{25}$$

$$CoM_y = \frac{m_h d_{h_y} + m_H d_{H_y} + m_{NVG} d_{NVG_y} - m_{CW} d_{CW_y}}{m_h + m_H + m_{NVG} + m_{CW}} \tag{26}$$

[8] This may explain why aircrew prefer to extend their neck and look under their NVGs to look at their cockpit instruments, rather than try to support their helmet system in a head down position as observed during familiarisation flights.

$$CoM_z = \frac{m_h d_{h_z} + m_H d_{H_z} + m_{NVG} d_{NVG_z} + m_{CW} d_{CW_z}}{m_h + m_H + m_{NVG} + m_{CW}} \tag{27}$$

Equation 25 implies that if the helmet system objects are centred along the yz-plane (x = 0), then all d_{o_x} are zero and therefore CoM_x is zero. From Equation 26, as m_{CW} increases, CoM_y decreases. Ideally, if $m_{CW} d_{CW_y} = m_H d_{H_y} + m_{NVG} d_{NVG_y}$ then the head and helmet system centre of mass in the y-direction is equal to the head centre of mass. Recall that this increase in CW is only beneficial to decreasing muscle torque when in a relatively static neutral posture. Otherwise the increasing m_{CW} will result in higher compression and shear forces as well as torques once in a non-neutral posture or joint accelerations are non-zero. Equation 27 suggests that increasing m_{CW} will move CoM_z even further away from the joint along the longitudinal axis, which has stability implications for the head/helmet system segment that is perched on the neck structure segment. Thus, tradeoffs exist between having a perfectly balanced helmet system using CW with the stability of the system as well as increased joint loads due to m_{CW}.

Summarising Section 3.1, relationships between C7 neck joint loads (dependent variables) were derived in terms of neck-supported mass properties and time-varying head postures (independent variables) using Newton's 2nd law of motion. The resultant physics-based model inferred basic solution design principles and guidelines such as 1) reduce total helmet system mass, 2) introduce low profile helmet systems, 3) promote biomechanically advantageous postures, 4) develop counterbalanced systems without using counter weights, and 5) fit helmets properly. All these potential solutions assume that reducing cumulative neck loads will reduce the risk of developing chronic pain.

3.2 Level of fidelity and 'fit-for-purpose'

Equations 8, 9, and 12 govern the compression force, shear force, and torque at a mass-less point at the C7 joint. These simple, low fidelity expressions seem to be sufficient for identifying solution principles and guidelines, even though there were a number of assumptions in deriving the equations.

If there is a need to relax the assumptions, then other more sophisticated (i.e., fewer simplifying assumptions) physics-based modelling techniques are available. For instance, if the mass-less neck joint assumption were relaxed, and neck joints have mass, size, and a simple shape, then Multi-Body Dynamics modelling can be used to determine the external neck loads at all seven neck joints as shown in Figure 4 (Fischer et al., 2014). This modelling technique was used to assess a new spring-activated counterbalance system that does not require CWs.

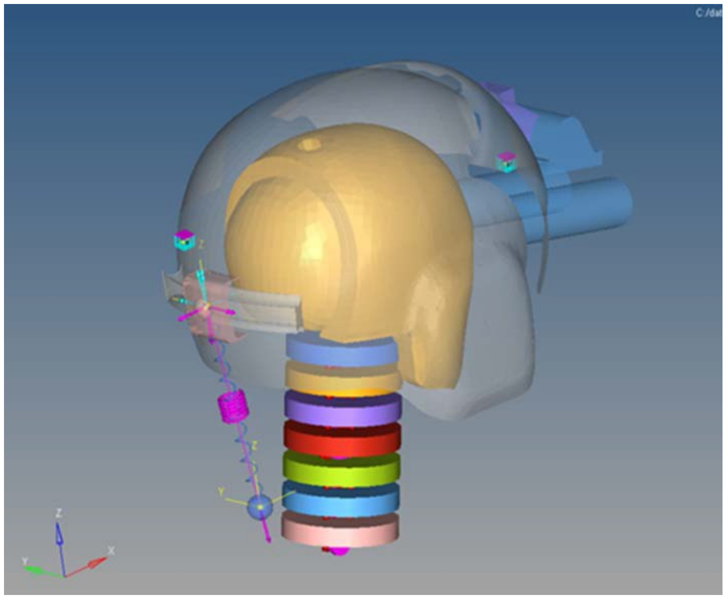

Figure 6: *Multi-body dynamics model that includes seven neck segments (Fischer et al., 2014).*

Moreover, finite element modelling was used to fully understand the cause and effect of neck loads on neck structure micro damage by calculating stresses and strains in neck muscles, ligaments, tendons, disks, and bones as shown in Figure 5 (Mustafy et al., 2014). Finite element models have an advantage of including experimentally derived hard and soft tissue constitutive material properties (i.e., empirical data); however, they often require long computational times to calculate the stresses and strains for each element in the model[9]. It becomes challenging to validate these models since sub-injurious data must be collected from post mortem human specimens of an age, size, and sex consistent with military populations and under military-relevant dynamic loading conditions. Such data are available in limited amounts, but while these provide bases for developing acute injury mechanisms and criteria, it would be difficult to relate these to chronic pain. Nevertheless, finite element models may be the only ethical and cost effective method to determine whether chronic neck pain is attributed to the cyclical loading of neck structures over a career.

These three modelling techniques—single joint, multi-body, and finite element models—have increasing levels of fidelity (along with decreasing number of simplifications); each with advantages and disadvantages. Thus, 'fit-for-purpose' must be considered when choosing a model for a specific application. For instance, a single joint model may be sufficient for the purpose of developing design principles and guidelines. A multi-body model may be sufficient assessing various solutions, such as new counterbalance systems. A finite element model may be used to develop an in-depth understanding of stress/strain relationships within and between neck structures to due external cumulative loading.

[9] Some studies focused on lumbar pain, (Schmidt, Paskoff, Shender, & Bass, 2012; Stemper, Baisden, Yoganandan, Shender, & Maiman, 2014) but little information is available related to how the effects of healing factor into predictions of stress and strain over time.

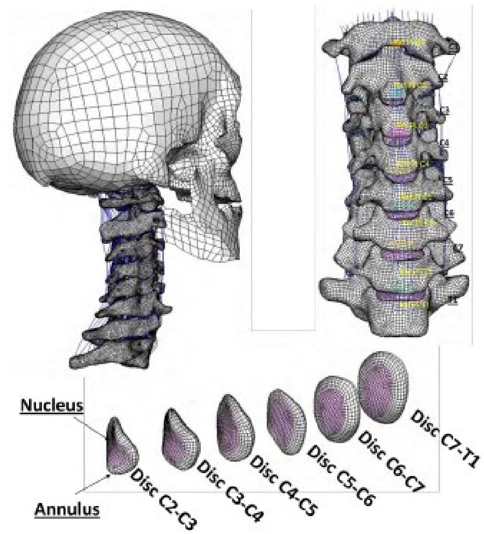

***Figure** 7: Finite element model using Abaqus Software (Moglo et al., 2012).*

3.3 Solution assessment using modelling and simulation

Recall that $\boldsymbol{a}$(t), $\boldsymbol{\alpha}$(t), and θ_i(t) vary with time as the aircraft accelerates and decelerates, and the head accelerates and decelerates relative to the torso as aircrew perform tasks. These neck load expressions may be incorporated in modelling and simulation (M&S) to determine the cumulative neck loading over time. We postulate that since neck loads during normal operations are well below acute injury thresholds (see Table 3), it is most likely cumulative neck load exposure over the course of a mission, multiple missions, and a career that cause chronic pain. Cumulative neck loads may be calculated (more precisely, estimated) using M&S with and without a mitigating solution to determine the extent to which the proposed solution reduces these loads.

As suggested in Section 3.2, multi-body models were used to assess solutions under a wide range of conditions before conducting the more expensive and time consuming human trials that often can only be done safely under a limited range of conditions (Tack & Nakaza, 2016). An M&S environment has been developed to test a number of proposed solutions aimed at minimising cumulative neck loads. The model is called the Integrated Mission, function, task analysis / Physical demands analysis Model (IMPM) which brings together both task and physical demands analyses to fully describe human work (Farrell, Tack, Nakaza, & Bray-Miners, 2016; Farrell, Tack, Nakaza, Bray-Miners, & Farrell, 2014; Tack et al., 2016; Tack & Nakaza, 2013).

At the heart of the IMPM modelling environment is a task and neck loads database. That is, CH-146 missions were decomposed into functions and tasks. Flying Pilot (FP), Non-Flying Pilot (NFP), and Flight Engineering (FE) tasks were further decomposed into postural sequences which, in turn, comprised of several individual postures. The linear acceleration was constant[10] ($\boldsymbol{a}$(t) = g) and $\boldsymbol{\alpha}$(t) and θ_i(t) for each task and postural sequence were measured at C7 using the XSenS™ motion capture system. The average compression force, shear force, and torque at C7 neck joint were estimated for each postural sequence

[10] Postural sequences were captured in a stationary helicopter and engine off; i.e., no motion or vibration. A feasibility study is underway to determine whether vibration can be added to the model.

using Visual 3D™: a multi-body dynamic model with realistic neck geometries and material properties. Neck loads were integrated across a mission by reconstituting tasks and postural sequences in the appropriate order, and normalising with respect to the total mission time.

IMPM included motion capture data for day- and night-helmet configurations (i.e., neck postures are different with and without NVGs). Neck loads were calculated for day and night missions, respectively. In almost every instance, the night missions yielded higher cumulative loads than the corresponding day condition. For some tasks during a night mission, aircrew look underneath their NVGs to read aircraft displays. Their necks are slightly extended and therefore the counterbalance has a positive effect in reducing muscle torque, compared to the equivalent day mission task with no NVGs and the neck is in a slightly flexed position. Compression and shear forces were always higher at night rather than during a day flight.

The following analyses were conducted using IMPM (Tack et al., 2014):

- Day and night average neck flexion/extension, axial rotation, and lateral bending angles were compared for FP, NFP, and FE. These angles were generally larger during night missions.
- Day and night neck torque, compression and shear forces were compared for each postural sequence and for FP, NFP, and FE. These loads were generally higher during night missions.
- Day and night cumulative neck torque, compression and shear forces were compared for Logistics Support and Surveillance Mission and Slung Load Training Mission, and for FP, NFP, and FE. Higher cumulative loads always occurred during night missions, and the training mission cumulative loads were generally higher than the logistics mission.
- IMPM was used as a Helmet System Mass Properties tool where mass and centre of mass were plotted against the average resultant torque across all postural sequences and for FP, NFP, and FE. As m increases and helmet system *CoM* moves away from the head *CoM*, the resultant torque increases as expected.

The following proposed solutions[11] have been assessed using IMPM (Bray-Miners, Osborne, & Tack, 2017; Tack et al., 2016):

- Biomechanically Advantageous Postures: M&S data indicated that some individuals employed alternative postural sequences that reduced neck loads (e.g., kneeling rather than lying prone to scan slung load). In some cases, neck loads were predicted to decrease by up to 85% by moving through a more efficient postural sequence. These biomechanically advantageous postures and tasks have been shown to reduce neck loads and this solution will be implemented by the CH146 Griffon Helicopter community (Farrell et al., 2017).
- FP and NFP MX-15 Task Sharing: a muscle fatigue model (Xia & Law, 2008) was added to IMPM to perform this assessment. The M&S results yielded an individual reduction of 71% muscle fatigue for the NFP when the tasks were shared as compared to when the NFP did the MX-15 task themselves. As expected, the FP muscle fatigue increased by 34%, when they shared the task; however, the combined NFP and FP neck muscle fatigue was reduced by 37% compared to the 'no task sharing' condition.
- Multi-Function Display (MFD): the MFD may include primary instruments, Control Display Unit, communications, maps, and radar altimeter functions. The MFD can be placed either on the front

[11] Additional motion capture data were collected for some of the proposed solutions.

panel, centre console, 'on knee', or 'optimal' (Figure 8) locations. These locations promote a more neutral posture. The assessment M&S showed that cumulative torque can be reduced by 35% for FP and NFP for both day and night flying for the optimal MFD location. But integrating an MFD into the current Griffon Helicopter would not be without significant costs.

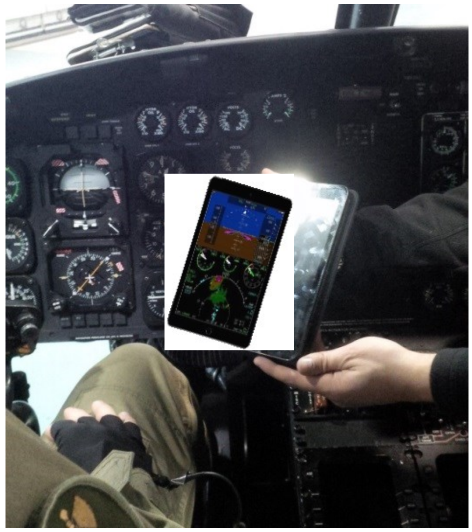

Figure 8: *Optimal MFD location.*

- Combined Ergonomic Solution: an optimal MFD with all the functions listed above plus ergonomics seats provided a more neutral posture for the Map/Doc Referencing, MX-15, CDU/Aircraft Management System, transit seated, scan seated, scan slung load, and scan regular landing tasks. IMPM was used to determine any differences in neck loading between this combined ergonomic solution and the current display and seating configuration. The FE resultant toque, for example, was reduced for both day and night logistics missions by 19% and 15% for logistics and training missions, respectively.

IMPM assessments show that solution combinations have the greatest potential for reducing neck loads, and in turn reducing the risk of chronic neck trouble. Note that at the heart of these assessments is multi-body physics-based model.

In this chapter, expressions were derived for the C7 neck joint loads using Newton's 2^{nd} law of motion. Solution design implications can be postulated from neck load expressions, including lighter, more balanced, and low profile helmets that reduce neck loads, counter weight and helmet fit design guidance, and solutions that promote a more upright position. These load equations form the basis for single-joint, multi-body, and finite element models (in order of increasing levels of fidelity). The choice of model should depend on the research objectives (i.e., 'fit-for-purpose'), such as understanding injury mechanisms, solution proposals, design, development, assessment, and advice.

The IMPM model was presented as a combination of, multi-body physics-based models, empirical data, and simulation used for solution assessment. Not all solutions can be tested in this environment. For example, it has been difficult to show how a solution that decreases neck loading also reduces the risk of chronic pain. Perhaps comparing measured values of neck loading to subjective ratings of pain levels may provide some insight into this relationship, which is the subject of empirical modelling in the next chapter.

4 Empirical models

While conceptual and physics-based models are constructed from logical argument and physical laws of nature, respectively, empirical models are derived from experimental measurements and observations. Typically, empirical models involve finding correlations between independent and dependent variables. If these correlations are statistically significant then causation might be inferred and a mathematical expression based on physics-based models may be developed that explains the statistical results. Otherwise, a mathematical equation still might be found that best fits the data, but the curve fit may not have any underlying meaning.

Empirical and physics-based models complement each other in that empirical models may help identify, refine, and even validate physics-based models, while physics-based models can explore conditions that cannot be conducted in an experimental setting due to resource constraints or safety concerns. The following sections provide examples of neck pain-related empirical models that complement the physics-based neck model equations in Chapter 3. Also presented is an example of empirical modelling where there is no physics-based model equivalent or it would be very difficult to generate such a model.

4.1 Correlating muscle activity with mass properties

Equations 8, 9, and 12 (neck load equations) describe a causal relationship between neck loads (dependent variables) that are derived from head-borne equipment mass properties, posture-related angles and accelerations, and time (independent variables). However, the equations do not indicate how these neck loads relate to acute or chronic neck pain. Neck muscles activate and recruit more muscles in response to additional neck loads. After some time, muscles begin to fatigue and are susceptible to overuse injury (Karakolis et al., 2015). Fatigued muscles and neck overuse injuries may trigger pain, but this possible neck pain mechanism is logical but extremely difficult to derive from physics-based modelling. Thus, empirical studies have been developed to determine any correlation between mass properties, neck loads, muscle activity, and subjective neck pain ratings. The following subsections provide brief descriptions of a few of these studies.

This first example of empirical modelling investigated whether there is an empirical relationship between muscle activity (muscle forces and torques) and neck loads. If there is a relationship, then one can reduce the number of variables, or degrees of freedom, that are related to chronic neck pain. As mentioned, the neck load equations indicate that as the helmet system m and MoI increase, and CoM moves further away from the head centre of mass, neck loads increase and therefore more muscle activity is required to move and stabilise the head and helmet system. Thus, from the equations we postulate that muscle activity varies directly with neck loads and mass properties. Data from the studies[12] were used to look for empirical evidence that support this postulate.

A neck supported mass study was conducted to determine the relationship between muscle activity, neck-supported mass properties and neck loading, and posture (Callaghan, 2014). The study objective was to determine whether muscle activity measured using surface electromyography (EMG) varied with different helmet configurations (i.e., different neck supported mass properties). Seven helmet configurations were tested from ‘no helmet’ to ‘helmet with NVGs down and CW’ (two most extreme conditions). Additionally

[12] A direct relationship with EMG analyses and reported pain has been inconsistent (Harrison et al., 2012).

for each helmet configuration, participants held seven separate flexion/extension, lateral bending, and rotation static postures for two minutes each.

"Overall, the wide variety of helmet configurations, which were tested under tightly controlled postural conditions, had minimal influence on the muscular [activity] during the low velocity movements and static holds examined in this study (Callaghan, 2014)." That is, muscle activation remained steady at approximately 5% of Maximum Voluntary Exertion (MVE) values[13] regardless of mass properties and neck loading, or posture. Thus, muscle activity did not seem to be correlated with neck loads or neck posture angles.

There were two possible explanations for this result. First, although deep muscle pairs mainly provide neck stability and control were likely recruited during the test runs, the EMG sensors measured the surface muscle groups' electric potential only. In order to determine muscle activity for deep muscles, fine EMG wires may be used (Burnett et al., 2007). These invasive techniques are uncomfortable and would be formidable to obtain ethics approval or recruit volunteers given that wire insertion would be near the spinal column. As neck loads increase, more muscle fibres are recruited to move and stabilize the neck. For sustained contractions, secondary muscles are recruited, muscle fatigue increases, and ligaments are recruited to provide additional support (Basmajian & Luca, 1985). If neck loads are high enough to produce injurious micro damage, then perhaps surface EMG can be used to objectively measure the muscle activation at these elevated and prolonged neck loads.

Fine EMG wire data taken from participants who may present with pain may be influenced by their reticence to perform movements or exertions that may exacerbate their discomfort. This makes classic EMG analysis difficult to interpret (Äng, Linder, & Harms-Ringdahl, 2005). Non-invasive techniques, such as magnetic resonance imaging (MRI) and quantitative computed tomography (QCT), have been used and can detect structural damage in clinically diagnosed radiculopathy patients (Nguyen et al., 2016). However, these techniques cannot be used as the load is applied.

Second, EMG signals rose during the initial and final high velocity phases (about five seconds long) as the study participants moved (accelerated and decelerated) their head to and from the required static posture. But EMG signals fell back to nominal levels during the two-minute static hold in the static posture. That is, muscle activity increased when the neck accelerated and decelerated from one head position to another, but remained low during static holds. This observation supports Equation 12 where muscle torque increases as the inertia term (i.e., angular acceleration) increases. Overall, this study found no correlation between muscle activity and static neck loads. However, elevated muscle activity was observed during brief dynamic neck movements.

A follow-on Neck Supported Mass study was conducted that measured muscle activation during realistic Griffon Helicopter aircrew tasks (McGuinness & Karakolis, 2016). The head-borne mass properties conditions included seven combinations of three different *m*, *CoM*, and *MoI* values. Preliminary results showed neck torque (calculated using motion capture data and Visual 3D™) varied with neck supported mass and centre of mass. The analysis also showed the expected correlation between muscle activity and changes in neck supported mass properties (i.e., neck loads).

A third study yielded a family of curves that related Cumulative Muscle Activation (CMA) to helmet system mass properties and aircrew role (Pilot and FE) (Farrell et al., in press).

[13] Sternocleidomastoid was the only muscle group that produced 10%–15% MVE.

The helmet systems included ANVIS 6 NVGs, a battery pack, and participant-specific Counter Weights attached to three helmets: In-Service HGU-56/P (IS), Improved HGU-56/P (IH), and Alpha Eagle (AE). *m*, *CoM*, and *MoI* were measured or calculated for each helmet system condition (IS, IH, and AE) and each participant (23 Pilots and 16 FEs). Given the median values for each helmet type and comparable size, IS was the heaviest, most unbalanced, and had the highest inertia values followed by AE, and then IH. According to the load equations in Chapter 3, we would expect that the IS helmet system would yield the highest muscle activity followed by AE and then IH.

In terms of aircrew roles, pilot behaviours generally featured more quasi-static tasks and erect postures and FE behaviours involved more dynamic neck movements and extreme postures. Given the observations of the previous studies and the load equations, it is expected that FEs would have higher muscle activity than pilots.

Figure 9 includes a family of curves that shows CMA as the dependent variable and helmet systems (mass properties) and roles (tasks and postures) as the two independent variables. Table 5 presents the Wilcoxon matched pairs test results for the helmet system pairs by role. For the FE role, there are no differences ($p > 0.05$) between the three helmet system types. For Pilot and All (FE and Pilot data combined) roles, there are no differences between AE and IH as well as IH and IS, but there is a significant difference between AE and IS ($p < 0.05$). Generally speaking, IS produced the same or higher CMA followed by AE and then IH regardless of roles as expected. Also as expected, the FE role consistently produced higher CMA than the Pilot role.

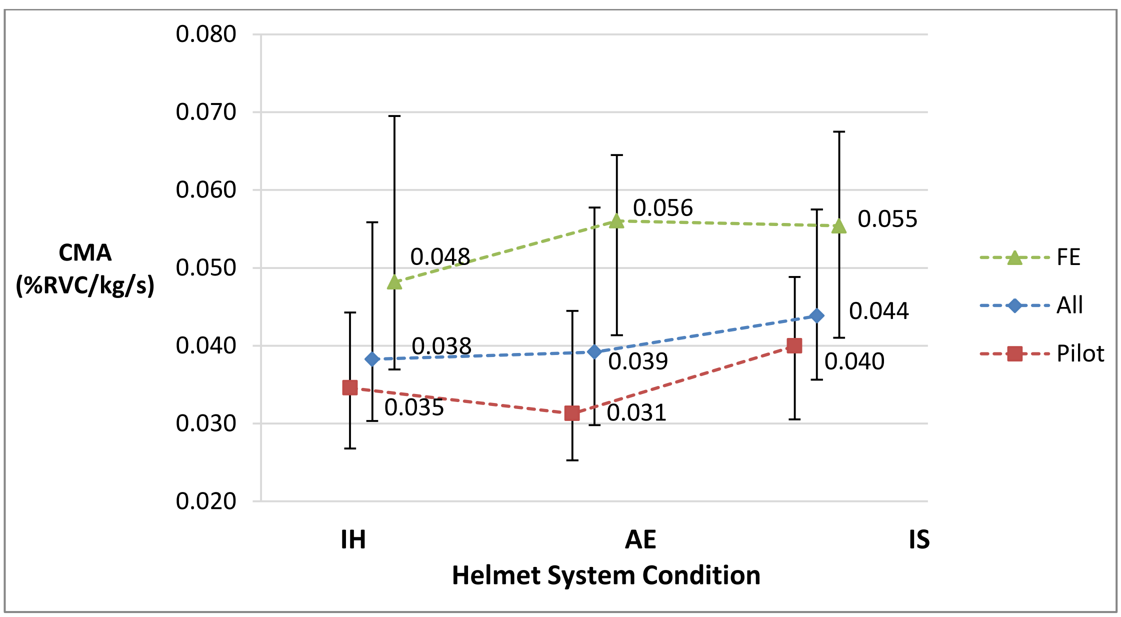

Figure 9: *Cumulative Muscle Activation (CMA) Medians and 25th to 75th percentile ranges. for FEs only, all participants, and Pilots only (Farrell et al., in press).*

Table 5: *Median (med) and 25^th^ to 75^th^ percentile range (25–75 %ile) for normalized CMA, followed. by Wilcoxon matched pairs test results for Helmet System pairs (Farrell et al., in press).*

Normalized CMA	**Role**	**IH**		**AE**		**IS**		**IH-AE**	**IH-IS**	**AE-IS**
		med	25-75 %ile	med	25-75 %ile	med	25-75 %ile	p	p	p
(%RVC/kg/s)	FE	0.048	0.037-0.069	0.056	0.041-0.064	0.055	0.041-0.068	0.363	0.140	0.438
	All	0.038	0.030-0.056	0.039	0.030-0.058	0.044	0.102-0.057	0.404	0.084	**0.005**
	Pilot	0.035	0.027-0.044	0.031	0.025-0.044	0.040	0.031-0.049	0.059	0.274	**0.002**

In summary, empirical data supports the hypothesis that muscle activity and neck loads are directly related to each other depending on the speed of movement and posture of aircrew tasks (i.e., aircrew role). Both the empirical and physics-based modelling results have revealed another independent variable that posture should be added to the conceptual model described by Equation 1 as follows:

$$\text{Chronic neck pain} = f_{new}\,(\text{exposure time, cumulative load, muscle activation, tissue damage, } \mathbf{posture}) \qquad (28)$$

'posture' refers to aircrew behaviours or tasks they perform while flying and their corresponding postural sequences that are made up of a series of postures and neck joint angles. Thus the neck trouble conceptual, physics-based, and empirical models are now consistent with each other.

4.2 Correlating neck pain with hot spots and groundings

In the previous example, an empirical model provided evidence for the physics-based relationship between muscle activity (force and torques) and helmet system mass properties as given in Equations 17, 18, and 23. In this section, empirical models are described that are formidable, if not impossible, to derive from physical or biological laws of nature.

The 2014 Griffon Helicopter Survey identified prevalence rates of musculoskeletal trouble amongst aircrew, and possible causal factors (Chafé & Farrell, 2016). A wide range of questions were administered from pain location, type, intensity, duration, and frequency to age and anthropometry, helmet mass and fit, aircrew roles and tasks, workspace and organisational issues. This allowed us to identify 'No Trouble (NT)' and 'Lifetime Prevalence of Trouble (NPT)' groups and look for any correlations between these groups and the five possible factors (see Figure 2). Most factors did not yield any significant differences between the two groups, except for hot spots, groundings, and preventive exercise.

Eleven questions were asked about Helmet Fit including hotspots, pressure points, looseness, strap tightness, movement, and overall fit at the time of the survey as well as at last fitting. A 7-point Likert scale was used where: 1was completely unacceptable and 7 was completely acceptable. The scores were ranked and Table 6 shows the results of a Mann-Whitney U test as well as Cohen's r calculation that provided insight into effect size. Benjamini-Hochberg correction for the total number of head-borne equipment-related tests yielded a corrected alpha value of $p \leq 0.008$. For neck and shoulder hot spots, there were significant differences between NT and LPT groups' mean rank score. That is, aircrew reporting trouble experience hot spots that are significantly less acceptable than aircrew with no trouble.

Table 6: *Mean rank score for helmet fit factors for groups reporting a lifetime prevalence of trouble and no trouble (Chafé & Farrell, 2016).*

Trouble location	**Mean rank score**		**Mann-Whitney U and Cohen's r**
Hot spots	**NT**	**LPT**	
Neck	124, n=5	100, n=160	U=3167, p=0.007; r=0.18
Low back	108, n=107	105, n=105	U=5430, p=0.642
Upper back	117, n=136	101, n=85	U=4974, p=0.052; r=0.13
Shoulder	115, n=135	93, n=78	U=4141, p=0.004; r=0.20
Pressure			
Neck	111, n=51	104, n=160	U=3831, p=0.474
Low back	105, n=107	108, n=105	U=5471, p=0.722
Upper back	112, n=136	109, n=85	U=5619, p=0.704
Shoulder	115, n=135	93, n=78	U=4198, p=0.007, r=0.18
Fit at last fitting			
Neck	111, n=51	104, n=159	U=3754, p=0.361
Low back	104, n=106	108, n=105	U=5351, p=0.580
Upper back	115, n=135	103, n=85	U=5066, p=0.093,r=0.11
Shoulder	115, n=135	92, n=77	U=4085, p=0.003, r=0.21

Inferences can be made from the simple empirical model with Helmet Fit as the independent variable and neck pain as the dependent variable. That is, a proper fit (that attempts to eliminate hot spots) is likely to reduce the risk of neck trouble, and vice versa.

Participants were asked to rate their pain intensity at the time of the survey as well as at the worst occurrence of pain. The possible responses on a 6-point scale included "0-no pain", "1-mild", "2-discomforting", "3-distressing", "4-horrible", and "5-excruciating". Participants also indicated whether they were ever grounded or 'benched' because of neck trouble. 'Benched' refers to an aircrew member who grounds themselves. The pain intensity was analysed for "not grounded" and "grounded" groups as well as "not benched" and "benched" groups. The worst pain intensity mean and standard deviation were found and Table 7 shows the results of an Independent t-test as well as Cohen's d calculation that provided insight into effect size. Benjamini-Hochberg correction for the total number of pain intensity-related tests yielded a corrected alpha value of $p \leq 0.041$. The worst pain intensity (> 3.1) was significantly higher for the 'grounded' and 'benched' groups than the 'not grounded' and 'not benched' groups (< 2.4) for neck, low back, upper back, and shoulder locations.

Inferences can be made from the simple empirical model with pain intensity as the independent variable and grounded or benched as the dependent variable. That is, a pain intensity greater than 3.1 (distressing) would likely lead to aircrew being grounded or benched. Solutions should be implemented so to reduce the pain intensity to below 2.4 (between discomfort and distressing) on the 6-point scale.

Table 7: *Mean worst pain intensity score ± stdev for grounded and benched participants (Chafé & Farrell, 2016).*

Trouble location	Mean worst pain intensity score ± stdev		Independent *t*-test and Cohen's d
	Not grounded	**Grounded**	
Neck	2.4 ± 0.7 (n=115)	3.2 ± 0.9 (n=37)	$t(150)=5.04$, $p<0.001$; d=0.96
Low back	2.4 ± 0.9 (n=44)	3.8 ± 0.9 (n=26)	$t(68)=6.42$, $p<0.001$; d=1.61
Upper back	2.3 ± 0.8 (n=46)	3.3 ± 1.1 (n=10)	$t(54)=3.17$, $p=0.003$; d=1.12
Shoulder	2.3 ± 0.8 (n=46)	3.3 ± 0.8 (n=11)	$t(55)=3.38$, $p=0.001$; d=1.15
	Not benched	**Benched**	
Neck	2.4 ± 0.8 (n=112)	3.1 ± 0.7 (n=40)	$t(150)=4.65$, $p<0.001$; d=0.86
Low back	2.4 ± 0.9 (n=44)	3.8 ± 0.9 (n=26)	$t(68)=6.42$, $p<0.001$; d=1.61
Upper back	2.3 ± 0.9 (n=47)	3.2 ± 1.0 (n=8)	$t(53)=2.55$, $p=0.014$; d=0.99
Shoulder	2.3 ± 0.8 (n=43)	3.1 ± 0.9 (n=13)	$t(54)=2.65$, $p=0.011$; d=0.85

In summary, Chapter 4 presents examples of empirical models that have independent and dependent variables, and a relationship between the variables from which one may make inferences. The first model that relates EMG to mass properties support the physics-based model described by Equations 17, 18, and 23, while the two models from survey data provide operationally relevant inferences that would be difficult, if not impossible, to derive from physical and biological laws of nature. On the other hand, empirical models are limited to only the population from which the data comes, and caution must be taken when using empirical models to generalise to other populations.

5 Conclusions

Aircrew neck-trouble models were used in this research for three key reasons. First, conceptual models were used to understand and frame the problem. Specifically, the NATO HFM 252 Neck Pain [conceptual] Framework included five factors: *Human* factors, aircrew *Behaviours*, body-borne *Equipment*, aircraft *Workspace*, and *Organisation* factors. This conceptual model also provided a common taxonomy for categorising solutions and generating testable hypotheses. This lead to a postulate in the form of an equation (Equation 28) stating that chronic neck pain is a function of exposure time, neck loads, muscle activity, tissue damage, and posture.

Second, physics-based models were used to find mathematical expressions of external neck loads as a function of time, neck-supported mass properties, and joint angles (Equations 17, 18, and 23). The neck load equations led to four key solution design guidance that reduce neck loads:

- Develop lighter, more balanced, and low profile helmets
- Wear minimal—if any—counter weights
- Ensure properly fitted helmet
- Use biomechanically advantageous postures

Single-joint, multi-body, and finite element physics-based models were presented in order of increasing levels of fidelity, and should be chosen depending on the research objectives. The single-joint model was chosen primarily to develop solution guidelines. Multi-body models were used to assess potential solutions. Finite element models have the potential to fully understand injury mechanisms and causation.

Third, empirical models were developed to show that muscle activity and neck loads are not related to each other for static postures, but are directly related to each other for dynamic movements. This model provided empirical evidence for the neck load equations. Also, empirical models were used to link neck pain to hotspots and groundings, which would be difficult to find this relationship with physic-based modelling. This analysis suggested that proper helmet fit should be considered to reduce hotspots, and other solutions should yield low pain scores (i.e., less than 2.4 on a 5-point pain scale) in order to avoid being grounded. Throughout these analyses, we conclude that models are essential for identifying knowledge gaps that are then used to determine when and how ethical and cost effective neck pain human volunteer experimental studies should be performed.

Future modelling efforts in this project include validating other variables in Equation 28, such as the relationship between muscle activity and tissue damage in predicting chronic neck pain, as well as better estimates for exposure time to episodic and chronic pain levels. Research is underway in determining aircrew performance decrements as a function of pain level.

The use of models in aircrew neck trouble research are essential tools for proposing, developing, assessing, and providing advice for neck pain mitigating solutions; that is, models are a means to an end and not an end in themselves. Although models may provide a deeper understanding of the aircrew neck pain problem, the ultimate objective is to use models to generate credible scientific advice on developing and validating potential mitigation solutions for the RCAF.

References

Äng, B. O. (2007). *Neck Pain in Air Force Pilots.* (Doctor of Philosophy), Karolinska Institutet, Stockholm, Sweden. (ISBN 978-91-7357-168-5).

Äng, B. O., Linder, J., & Harms-Ringdahl, K. (2005). Neck/Shoulder Exercise for Neck Pain in Air Force Helicopter Pilots: A Randomized Controlled Trial. *Aviation, Space, and Environmental Medicine, 76*(4), 375–380.

Äng, B. O., Monnier, A., & Harms-Ringdahl, K. (2009). Neck/Shoulder Exercise for Neck Pain in Air Force Helicopter Pilots: A Randomized Controlled Trial. *Spine, 34*(16), E544–E551.

Basmajian, J. V., & Luca, C. J. D. (1985). *Muscles Alive: Their functions revealed by Electromyography* (5 ed.). Philadelphia: Lippincott Williams and Wilkins.

Blackmore, M. E., Goswami, T., & Chancey, C. (2012). Cervical Spinal Injuries and Risk Assessment. In T. Goswami (Ed.), *Injury and Skeletal Biomechanics* (pp. 18). online: InTech.

Bowden, T. (1987). Back Pain in Helicopter Aircrew: A Literature Review. *Aviation, Space, and Environmental Medicine, 58*, 461–467.

Bowden, T. J. (1984). Backache in Helicopter Aircrew, The current Situation: A Literature Review (pp. 50). Downsview, Ontario, Canada: Technical Report DCIEM No. 84-R-39 Defence and Civil Institute of Environmental Medicine.

Bray-Miners, J., Osborne, A., & Tack, D. W. (2017). Modeling Assessment of Crewstation ergonomic Solutions (pp. 76). Scientific Authority: Giovanni Fusina, Ph.D., Contract Report, DRDC-RDDC-2017-C140, Defence Research and Development Canada.

Burnett, A., Green, J., Netto, K., & Rodrigues, J. (2007). Examination of EMG normalisation methods for the study of the posterior and posterolateral neck muscles in healthy controls. *Journal of Electromyography and Kinesiology, 17*(5), 635–641. doi: https://doi.org/10.1016/j.jelekin.2006.06.003.

Callaghan, J. P. (2014). The Influence of Neck Posture and Helmet Configuration on Neck Muscle Demands (pp. 125). Scientific Authority: Ann Nakashima, Contract Report, DRDC-RDDC-2014-C188, Defence Research and Development Canada.

Chafe, G. S. (2013). Evaluation of Potential Solutions to Mitigate Aircrew Neck Strain *CIMVHR contract W7714-125624/001/SV-task 14* (pp. 64). Toronto, Canada: Technical Report, DRDC Toronto TR 2013-089, Defence Research and Development Canada.

Chafé, G. S., & Farrell, P. S. E. (2016). Royal Canadian Air Force CH-146 Griffon aircrew 2014 spinal musculoskeletal trouble survey. Scientific Report, DRDC-RDDC-2016-R179, Defence Research and Development Canada.

Chen, Y., Wickramasinghe, V., & Zimcik, D. G. (2011). Development of Adaptive Helicopter Seat for Aircrew Vibration Reduction. *Journal of Intelligent Material Systems and Structures, 22*(5), 489–502. doi: 10.1177/1045389x11400343.

Dibblee, J., Worthy, P., Farrell, P. S. E., Hetzler, M., Reid, S., Stevenson, J., & Fischer, S. (2015). Evaluating a prototype device designed to alleviate night vision goggle induced neck strain among military personnel. *Ergonomics, 58*(12), 2067–2077.

DND/CAF. (2013). *Royal Canadian Air Force Future Concepts Directive*. (RCAF FCD 1.0,). Ottawa, Canada.

Farrell, P. S. E., Chafé, G. S., Karakolis, T., Fusina, G., Hollands, J. G., & Wright-Beatty, H. (2017). Interim scientific advice for six RCAF Griffon helicopter aircrew neck-trouble mitigating solutions. Scientific Letter, DRDC-RDDC-2016-L028, Defence Research and Development Canada.

Farrell, P. S. E., Goff, C. P., Shender, B. S., Chancey, V. C., Crowley, J., Day, S. E., . . . Wright-Beatty, H. (draft). Aircrew Neck Pain Prevention and Management. Task Group HFM-252 Final Report: Collaboration Support Office - NATO S&T Organization, Paris, France.

Farrell, P. S. E., Maceda, E., Niri, A., Apostoli, C., McGuinness, C., Vallikanthan, J., & Karakolis, T. (in press). Helmet Systems Comparisons using Neck Muscle Activity and External Load measurements: Helmet Systems Assessment Trial (pp. 28): Scientific Report, Defence Research and Development Canada.

Farrell, P. S. E., Tack, D. W., Nakaza, E. T., & Bray-Miners, J. (2016). *Helicopter Aircrew Cumulative Neck Loads from Integrated Task and Physical Demands Analyses*. Paper presented at the HFES 2016 Annual Meeting, Washington, DC.

Farrell, P. S. E., Tack, D. W., Nakaza, E. T., Bray-Miners, J., & Farrell, C. M. (2014). *Combined Task and Physical Demands Analyses towards a Comprehensive Human Work Model*. Paper presented at the HFES 2014 Annual Meeting, Chicago, Il.

Fernie, G., & Mayich, J. (2013). Mitigating Neck Pain in Aircrew *A Proposal for the Development of Near-Term Solutions* (pp. 74). Toronto, Canada: Scientific Authority: Capt. Gabrielle S. Chafe, Contract Report, DRDC Toronto CR 2013-038, Defence Research and Development Canada.

Fischer, S. L., Stevenson, J. M., Albert, W. J., Harrison, M. F., Bryant, T., Beach, T. A. C., . . . Coffey, B. (2013). Near-Term Ideas to Address Aircrew Helmet Systems-Induced Neck Pain *Mitigating Neck Pain in Aircrew* (pp. 85). Toronto, Canada: Scientific Authority: Capt. Gabrielle S. Chafe, Contract Report, DRDC Toronto CR 2013-039, Defence Research and Development Canada.

Fischer, S. L., Stevenson, J. M., Reid, S. A., & Hetzer, M. (2014). Spring-Based Helmet System Support Prototype to Address Aircrew Neck Strain (pp. 95). Scientific Authority: Philip S. E. Farrell, Contract Report. DRDC-RDDC-2014-C187, Defence Research and Development Canada.

Forde, K. A., Albert, W. J., Harrison, M. F., Neary, J. P., Croll, J., & Callaghan, J. P. (2011). Neck loads and posture exposure of helicopter pilots during simulated day and night flights. *International Journal of Industrial Ergonomics, 41*(2011), 128–135.

Fraser, W. D., Crowley, J. S., Shender, B. S., & Lee, V. M. (2015). Multinational survey of neck pain in rotor-wing aircrew (pp. 24). TTCP Technical Report DOC-HUM-01-2015: The Technical Cooperation Program.

Guzman, J., Hurwitz, E. L., Carroll, L. J., Haldeman, S., Côté, P., Carragee, E. J., . . . Cassidy, J. D. (2009). A new conceptual model of neck pain: Linking onset, course, and care: the Bone and Joint Decade 2000-2010 Task Force on Neck Pain and Its Associated Disorders. *Journal of Manipulative and Physiological Therapeutics, 32*(2), S17–S28. doi: http://dx.doi.org/10.1016/j.jmpt.2008.11.007.

Harrison, M. F., Coffey, B., Albert, W. J., & Fischer, S. L. (2015). Night Vision Goggle-Induced Neck Pain in Military Helicopter Aircrew: A Literature Review. *Aerospace Medicine and Human Performance, 86*(1), 46–55.

Harrison, M. F., Neary, J. P., Albert, W. J., & Croll, J. C. (2012). A Predictive Logistic Regession Equation for Neck Pain in Helicopter Aircrew. *Aviation, Space, and Environmental Medicine, 83*(6), 604–608.

Hawes, R., Whitehead, J., & Gray, G. (2014). *Canadian Forces Health Examinations and Record outcomes (CF-HERO) Database: Primary Care Diagnoses of Neck & Back Pain in Royal Canadian Air Force (RCAF) Aircrew*. Power Point Presentation. Department of National Defence. Canadian Forces Health Services Epidemiology Section, Directorate of Force Health Protection.

Hébert, L. J., Roy, J.-S., Burke, P., Côté, J. N., & Grodecki, R. (2012). *Validation of a Standardized Neck/Shoulder Exercise Program in CF Helicopter Aircrew and its Impact on Neck Pain, Impairments and Functional Limitations/Restrictions*. Project for synopsis # 2012-07-038-001-0003 approved by the SGHRB.

Howarth, S. J., & Callaghan, J. P. (2013). Towards establishing an occupational threshold for cumulative shear force in the vertebral joint — An in vitro evaluation of a risk factor for spondylolytic fractures using porcine specimens. *Clinical Biomechanics, 28*(3), 246–254. doi: http://dx.doi.org/10.1016/j.clinbiomech.2013.01.003.

Howarth, S. J., Karakolis, T., & Callaghan, J. P. (2015). A finite element evaluation of the moment arm hypothesis for altered vertebral shear failure force. *Computer Methods in Biomechanics and Biomedical Engineering, 18*(5), 545–555. doi: http://dx.doi.org/10.1080/10255842.2013.820717.

Jahanshah, F., Fan, C., Lu, Z., Kapps, A., Yen, L., Zhao, S., . . . Rebello, J. (2006). Development, Design and Integration of Neck Support Systems for Helicopter Pilots (pp. 152). Scientific Authority: William D. Fraser, Contract Report, DRDC-RDDC-2018-C021, Defence Research and Development Canada.

Jahanshah, F., Krzeminski, M., Lu, Z., Ma, A., Kapps, A., Aly, A., & Li, R. (2005). Development, Design and Integration of Neck Support Systems for S&R Helicopter Pilots (pp. 162). Scientific Authority: William D. Fraser, Contract Report, DRDC-RDDC-2018-C020, Defence Research and Development Canada.

Karakolis, T., Farrell, P., & Fusina, G. (2015). Neck overuse injury in CH-146 Griffon helicopter aircrews. *Procedia Manufacturing.*, 3:4205–4212. doi: 10.1016/j.promfg.2015.07.396.

Karakolis, T., McGuinness, C., & Farrell, P. S. E. (in press). Neck Supported Mass Study Report (pp. 107): Scientific Report, Defence Research and Development Canada.

Kingma, I., de Looze, M. P., Toussaint, H. M., Klijnsma, H. G., & Bruijnen, T. B. M. (1996). Validation of a full body 3-D dynamic linked segment model. *Human Movement Science, 15*(1996), 833–860.

Maksel, R. (2014). The Helicopter Pilot's Dreaded Helo Hunch: Flying helicopters is a musculoskeletal nightmare. Retrieved 27 August 2015, from http://www.airspacemag.com/daily-planet/dreaded-helo-hunch-180950037/.

Manoogian, S. J., Kennedy, E. A., & Duma, S. M. (2005). A Literature Review of Musculoskeletal Injuries to the Human Neck and the Effects of Head-Supported Mass Worn by Soldiers (pp. 58). Fort Belvoir, VA: USAARL Contract Report No. CR-2006-01, Defense Technical Information Center, Fort Belvoir, Virginia 22060-6218.

McGill, S. M. (2007). *Low back disorders: Evidence based prevention and rehabilitation* (Second ed.). Champaign, IL, U.S.A.: Human Kinetics Publishers.

McGuinness, C., & Karakolis, T. (2016). *Effect of Head Mounted Mass Properties on Neck Joint Reaction Torques While Performing Simulated Helicopter Pilot Tasks*. Paper presented at the 19th Biennial Meeting of the Canadian Society for Biomechanics (CSB 2016), Hamilton, Ontario.

McLaughlin, T. (2013). Neck Strain in Air Crew Literature Search *STI Search 15458* (pp. 74). Ottawa, Canada: DND Project Manager: Maj. Graham Fisher, Contract Report, Human System Integration/Human Factors Engineering, Directorate of Technical Airworthiness and Engineering Support, Department of National Defence, Canada.

Meriam, J. L. (1980). *Engineering Mechanics Statics and Dynamics*. Toronto: John Wiley & Sons.

Moglo, K., Mesfar, W., & Mustafy, T. (2012). Finite Element Model of the Human Cervical Spine for Evaluation of Head and Neck Injuries (pp. 16). Kingston, ON, Canada: Project Manager: Ms Tara Foster-Hunt and Capt. Andrea MacEachern, Mechanical Engineering Technical Report #120901, Directorate Aerospace Engineering Support (DTAES 6).

Moroney, S. P., Schultz, A. B., & Miller, J. A. A. (1988). Analysis and Measurement of Neck Loads. *Journal of Orthopaedic Research, 6*(5), 713–720, RAven Press, Ltd., New York.

Mustafy, T., El-Rich, M., Mesar, W., & Moglo, K. (2014). Investigation of impact loading rate effects on the ligamentous cervical spinal load-partitioning using finite element model of functional spinal unit C2–C3. *Journal of Biomechanics, 47*, 2891–2903.

NATO. (2013). Terms of Reference (ToR): HFM-ET-126 Approval for Transition to TG on Aircrew Neck Pain. Terms of Reference HFM 252: Collaboration Support Office - NATO S&T Organization, Paris, France.

Nguyen, H. S., Doan, N., Shabani, S., Baisden, J., Wolfla, C., Paskoff, G., . . . Stemper, B. (2016). Upright magnetic resonance imaging of the lumbar spine: Back pain and radiculopathy. *Journal of Craniovertebral Junction and Spine, 7*(7), 31–37.

Parkinson, R. J., & Callaghan, J. P. (2007). Can periods of static loading be used to enhance the resistance of the spine to cumulative compression? *Journal of Biomechanics, 40*(13), 2944–2952.

Pollak, K. A., Swenson, J. D., Vanhaitsma, T. A., Hughen, R., Jo, D., Light, K. C., . . . Light, A. R. (2014). Exogenously applied muscle metabolites synergistically evoke sensations of muscle fatigue and pain in human subjects. *Experimental Physiology, 99*(2), 368–380. doi: 10.1113/expphysiol.2013.075812.

Schmidt, A. L., Paskoff, G., Shender, B. S., & Bass, C. R. (2012). Risk of Lumbar Spine Injury from Cyclic Axial Loading. *Spine, 37*(26), E1614–1621. doi: 10.1097/BRS.0b013e3182752a19.

Shanahan, D. F., & Reading, T. E. (1984). Helicopter Pilot Back Pain: A Preliminary Study. *Aviation Space and Environmental Medicine, 55*(2), 117–121.

Stemper, B. D., Baisden, J. L., Yoganandan, N., Shender, B. S., & Maiman, D. J. (2014). Mechanical Yield of the Lumbar Annulus: A Possible Contributor to Instability. *Journal of Neurosurgery. Spine, Oct*(4), 608–613. doi: 10.3171/2014.6.SPINE13401.

Tack, D. W., Bray-Miners, J., Nakaza, E. T., & Osborne, A. (2016). Griffon 2 Project: Assessment of Revised Work Distribution for Griffon Helicopter Operator (pp. 71). Scientific Authority: Philip S. E. Farrell, Ph.D., Contract Report, DRDC-RDDC-2016-C224, Defence Research and Development Canada.

Tack, D. W., Bray-Miners, J., Nakaza, E. T., Osborne, A., & Mangan, B. (2014). Griffon Helicopter Neck Strain Project. Part 1: Mission Function Task Analysis and Physical Demands Analysis Report. Part 2: Physical Demands Analysis Library. (pp. 340). Scientific Authority: Philip S. E. Farrell, Ph.D., Contract Report, DRDC-RDDC-2014-C228, Defence Research and Development Canada.

Tack, D. W., & Nakaza, E. T. (2013). Mission, Function, Task Analysis / Physical Demands Analysis Plan for Griffon Helicopter Crew Neck Strain Project (pp. 36). Toronto, Canada: Scientific Authority: Philip S. E. Farrell, Ph.D., Contract Report, DRDC Toronto CR 2013-139, Defence Research and Development Canada.

Tack, D. W., & Nakaza, E. T. (2016). Griffon 2 Project - Solution Space Workshop: Rating of Neck Trouble Mitigation Concepts for Griffon Helicopter Aircrew (pp. 109). Scientific Authority: Philip S. E. Farrell, Ph.D., Contract Report, DRDC-RDDC-2016-C218, Defence Research and Development Canada.

Tampier, C., Drake, J. D., Callaghan, J. P., & McGill, S. M. (2007). Progressive disc herniation: an investigation of the mechanism using radiologic, histochemical, and microscopic dissection techniques on a porcine model. *Spine (Phila Pa 1976), 32*(25), 2869–2874. doi: 10.1097/BRS.0b013e31815b64f5.

van den Oord, M. H. A. H. (2012). *Prevention of flight-related neck pain in military aircrew.* (Doctor of Philosophy), University of Amsterdam, Amsterdam, Netherlands. (ISBN 978-90-804423-0-6).

Vette, A. H., Yoshida, T., Thrasher, T. A., Masani, M., & Popovic, M. R. (2012). A comprehensive three-dimensional dynamic model of the human head and trunk for estimating lumbar and cervical joint torques and forces from upper body kinematics. *Medical Engineering & Physics, 34*, 640–649.

Xia, T., & Law, L. A. F. (2008). A theoretical approach for modeling peripheral muscle fatigue and recovery. *Journal of Biomechanics, 41*, 3046–3052.

Xiao, A., & Farrell, P. S. E. (2016). Royal Canadian Air Force Neck- and Back-trouble Research: A Historical Review: Reference Document, DRDC-RDDC-2016-D031, Defence Research and Development Canada.

List of symbols/abbreviations/acronyms/initialisms

AE	Alpha Eagle helmet
avg	average
CAF	Canadian Armed Forces
CFEME	Canadian Forces Environmental Medicine Establishment
CFHS	Canadian Forces Health Services
ch	chronic
cm	centimetre
CMA	Cumulative Muscle Activation
CoM	Centre of Mass
CoP	Community of Practice
CP	Collaborative Project
CW	Counter Weight
DND	Department of National Defence
DRDC	Defence Research and Development Canada
DTAES	Director of Technical Airworthiness and Engineering Support
EMG	Electromyography
ep	episodic
FE	Flight Engineer
FP	Flying Pilot
HFM	Human Factors and Medicine
HMD	Helmet Mounted Display
IH	Improved HGU 56/P helmet
IMPM	Integrated Mission function task analysis / Physical demands analysis Model
IS	In-Service helmet
kg	kilogram
m	mass
M&S	Modelling and Simulation
MFD	Multi-Function Display
MoI	Mass Moment of Inertia
MRI	Magnetic Resonance Imaging

MVE	Maximum Voluntary Exertion
n	number of participants
NATO	North Atlantic Treaty Organisation
nc	non-chronic
NFP	Non-Flying Pilot
NVG	Night Vision Googles
QCT	Quantitative Computer Tomography
RCAF	Royal Canadian Air Force
RTO	Research and Technology Organisation
std	standard deviation
TP	Technical Panel
TTCP	The Technical Cooperation Program
WBE	Work Breakdown Element

DOCUMENT CONTROL DATA

(Security markings for the title, abstract and indexing annotation must be entered when the document is Classified or Designated)

1. ORIGINATOR (The name and address of the organization preparing the document. Organizations for whom the document was prepared, e.g., Centre sponsoring a contractor's report, or tasking agency, are entered in Section 8.) DRDC – Toronto Research Centre Defence Research and Development Canada 1133 Sheppard Avenue West P.O. Box 2000 Toronto, Ontario M3M 3B9 Canada	2a. SECURITY MARKING (Overall security marking of the document including special supplemental markings if applicable.) CAN UNCLASSIFIED 2b. CONTROLLED GOODS NON-CONTROLLED GOODS DMC A

3. TITLE (The complete document title as indicated on the title page. Its classification should be indicated by the appropriate abbreviation (S, C or U) in parentheses after the title.)

The Use of Models in Aircrew Neck Trouble Research

4. AUTHORS (last name, followed by initials – ranks, titles, etc., not to be used)

Farrell, P.S.E.; Shender, B.S.; Fusina, G.

5. DATE OF PUBLICATION (Month and year of publication of document.)	6a. NO. OF PAGES (Total containing information, including Annexes, Appendices, etc.)	6b. NO. OF REFS (Total cited in document.)
July 2018	42	62

7. DESCRIPTIVE NOTES (The category of the document, e.g., technical report, technical note or memorandum. If appropriate, enter the type of report, e.g., interim, progress, summary, annual or final. Give the inclusive dates when a specific reporting period is covered.)

Scientific Report

8. SPONSORING ACTIVITY (The name of the department project office or laboratory sponsoring the research and development – include address.)

DRDC – Toronto Research Centre
Defence Research and Development Canada
1133 Sheppard Avenue West
P.O. Box 2000
Toronto, Ontario M3M 3B9
Canada

9a. PROJECT OR GRANT NO. (If appropriate, the applicable research and development project or grant number under which the document was written. Please specify whether project or grant.)	9b. CONTRACT NO. (If appropriate, the applicable number under which the document was written.)
10a. ORIGINATOR'S DOCUMENT NUMBER (The official document number by which the document is identified by the originating activity. This number must be unique to this document.) DRDC-RDDC-2018-R147	10b. OTHER DOCUMENT NO(s). (Any other numbers which may be assigned this document either by the originator or by the sponsor.)

11a. FUTURE DISTRIBUTION (Any limitations on further dissemination of the document, other than those imposed by security classification.)

Public release

11b. FUTURE DISTRIBUTION OUTSIDE CANADA (Any limitations on further dissemination of the document, other than those imposed by security classification.)

12. ABSTRACT (A brief and factual summary of the document. It may also appear elsewhere in the body of the document itself. It is highly desirable that the abstract of classified documents be unclassified. Each paragraph of the abstract shall begin with an indication of the security classification of the information in the paragraph (unless the document itself is unclassified) represented as (S), (C), (R), or (U). It is not necessary to include here abstracts in both official languages unless the text is bilingual.)

Seventy-five percent of Griffon Helicopter aircrew have reported various levels of neck pain as a result of performing flying tasks, and 23% have been grounded due to neck trouble. In the past, Defence Research and Development Canada (DRDC) has been involved in neck trouble research focussing on Finite Element modelling techniques in order to understand neck pain mechanisms. Some have argued that this line of research has not led to tangible solutions in Canada; however, others have used modelling as a necessary tool in the development of neck pain mitigating solutions. This paper argues that conceptual, physics based, and empirical models can be used at various stages to help define the scope of the problem, suitable verification and validation tests and provide a path for the solution's development, design, and assessment. This has led to a description of the NATO HFM 252 framework as well as mathematical equations that estimate neck loading due to helmet system mass properties. It is concluded that these modelling activities are essential for ethical and cost effective neck trouble research.

Soixante-quinze pour cent des membres d'équipage des hélicoptères Griffon ont signalé divers niveaux de douleur au cou à la suite de l'exécution de tâches en vol, et 23 % ont été cloués au sol en raison de maux de cou. Par le passé, Recherche et développement pour la défense Canada (RDDC) a participé à des recherches sur les maux de cou axées sur les techniques de modélisation par éléments finis afin de comprendre les mécanismes de la douleur cervicale. Certains ont soutenu que cet axe de recherche n'a pas abouti à des solutions tangibles au Canada. Cependant, d'autres ont utilisé la modélisation comme outil nécessaire à l'élaboration de solutions pour atténuer les douleurs au cou. Dans ce document, on soutient que les modèles conceptuels, physiques et empiriques peuvent être utilisés à diverses étapes pour aider à définir la portée du problème, de même que les tests de vérification et de validation appropriés, ainsi qu'à offrir une avenue pour l'élaboration, la conception et l'évaluation d'une solution. Cela a débouché sur une description du cadre HFM 252 de l'OTAN, ainsi que sur des équations mathématiques permettant d'évaluer la charge cervicale en fonction des caractéristiques de masse du système de casque. On en conclut que ces activités de modélisation sont essentielles à une recherche éthique et rentable sur les maux de cou.

13. KEYWORDS, DESCRIPTORS or IDENTIFIERS (Technically meaningful terms or short phrases that characterize a document and could be helpful in cataloguing the document. They should be selected so that no security classification is required. Identifiers, such as equipment model designation, trade name, military project code name, geographic location may also be included. If possible keywords should be selected from a published thesaurus, e.g., Thesaurus of Engineering and Scientific Terms (TEST) and that thesaurus identified. If it is not possible to select indexing terms which are Unclassified, the classification of each should be indicated as with the title.)

Muscle Activation; Neck Load; neck pain; tissue damage; exposure time; body-borne equipment; aircrew behaviour; tasks; postures; aircraft workspace; chronic; episodic; non-chronic