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ON NECK LOAD AMONG HELICOPTER PILOTS

EFFECTS OF HEAD-WORN EQUIPMENT, WHOLE-BODY VIBRATION AND NECK POSITION

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To Sara & Theodor

ABSTRACT

Introduction: Helicopter pilots complain of pain originating from the neck region. The causes are still basically unknown, but the ergonomic situation when flying a helicopter, with unfavorable load caused by static neck and body positions, whole-body vibration and heavy head-worn equipment, has been suggested as a risk factor.
Aim: The aim of the work reported in this thesis was to quantify the effects of external loads on helicopter pilots' necks, and to evaluate different methods for measuring neck load.
Methods: Thirty-nine Swedish military helicopter pilots participated in the five studies. The effects of different neck and body positions, head-worn equipment and vertical whole-body vibration were evaluated concerning neck muscle activity, induced mechanical load and seat-to-head transmissibility. Surface electromyograms (EMG) were recorded from upper and lower dorsal neck muscles, the sternocleidomastoid, and the upper trapezius. The induced load on the lower cervical spine was calculated using a sagittal, static, biomechanical model, and vibration transmissibility was calculated as the ratio of recordings from helmet-mounted accelerometers and vertical vibration acceleration measured at the seat. The neck and body positions evaluated were: neutral, neck flexion 20° (for muscle activity, induced load and transmissibility), neck rotation 30° (EMG), and trunk inclination 20° (EMG). The head-worn equipment evaluated was: helmet alone, helmet and night vision goggles (NVG), and helmet, NVG and counterweight (all evaluated using EMG, induced load and transmissibility). Vibration was evaluated at different frequencies (2.5-30 Hz) and magnitudes (0.5, 1, and 2 m/s^2) using EMG and transmissibility.
For the reliability testing of a neck fatigue protocol, the pilots performed isometric contractions in neck flexion and extension for 45 s, sustaining a force representing 75 % of maximum strength in a seated position. Subjective fatigue was rated using the Borg CR-10 scale. The test was repeated twice the first day and then two additional times with one-week intervals. Variables analyzed were the slope of the median frequency change, the normalized slope, and the ratings after 15, 30 and 45 s; and also the initial median frequency (IMDF). The intra-class correlation (ICC) and the measurement error (S_w), intra- and inter-day were calculated.
Results: Dorsal neck muscle activity increased by 3-4 % of maximum voluntary electrical activation (MVE) as a cause of neck rotation, 2-3 % of neck flexion, and 1.5-2.5 % of trunk inclination. The use of NVG increased muscle activity in upper neck by 0.5-1.5 % and in lower neck by about 0.5 %. Results with added counterweight were about the same as with NVG. Muscle activity increased by about 0.5-1 % MVE as a function of vibration at frequencies around 4-5 Hz, with the higher levels when the neck was flexed. Muscle activity was also affected by vibration magnitude, where lower-neck-muscle and trapezius activity increased at the highest vibration level at frequencies around 4-5 Hz. The induced load was also affected by both neck flexion and NVG. The load at 20° flexion increased by about 8 % of maximum voluntary contraction (MVC) compared to neutral and by about 3 % MVC when adding NVG compared to using helmet only. The load decreased somewhat when a counterweight was added.
The transmissibility peak in a vertical direction was highest when the head was in neutral position and the fore-and-aft transmissibility peak was highest when the head was flexed. There were no effects of head-worn equipment concerning vertical transmissibility, but the fore-and-aft transmissibility peak level was lower with NVG. Different magnitudes of vibration gave only minor effects on transmissibility.
The best reliability for the slope was found for the 45 s intra-day analysis, taking all measurements into account (ICC 0.65-0.83). Reliability after 30 s was poorer but still acceptable (ICC 0.52-0.71). For the subjective ratings, the highest reliability was found after 30 s inter-day (ICC 0.86-0.88). IMDF showed generally high reliability for the intra-day analyses (ICC 0.63-0.80).
Conclusion: All three proposed risk factors caused measurable changes in muscle activity, induced load and seat-to-head transmissibility. Of the three, neck and body position caused the highest response.
EMG and seat-to-head transmissibility responded somewhat different as function of vibration indicating that effects of vibration should be measured using more then one outcome measure.
The protocol for measuring neck muscle fatigue can be considered reliable for use in further research. Since performing a contraction of 75 % of maximum was quite strenuous, it is recommended that the protocol period be shortened to 30 s.

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LIST OF PUBLICATIONS

The thesis is based on the following publications, which are referred to in the following text by their Roman numerals. Some new analyses and results not been presented elsewhere have also been added.

I	Thuresson, M., Äng, B., Linder, J., and Harms-Ringdahl, K. Neck muscle activity in helicopter pilots: effect of position and helmet-mounted equipment. Aviation, Space & Environmental Medicine. 74, 527-532, 2003.
II	Thuresson, M., Äng, B., Linder, J., and Harms-Ringdahl, K. Mechanical load and EMG activity in the neck induced by different head-worn equipment and neck postures. International Journal of Industrial Ergonomics. 35:13-18, 2005.
III	Thuresson, M., Wallin, H.P., and Harms-Ringdahl, K. Effect of vertical sinusoidal vibration exposure, posture and helmet-mounted equipment on neck muscle activity and seat-to-head transmissibility among helicopter pilots. Submitted 2005
IV	Thuresson, M., and Wallin, H.P. Effects of different magnitudes of vertical whole-body vibration on neck muscle activity and seat-to-head transmissibility among helicopter pilots. Manuscript 2005
V	Thuresson, M., Äng, B., Linder, J., and Harms-Ringdahl, K. Intra-rater reliability of electromyographic recordings and subjective evaluation of neck muscle fatigue among helicopter pilots. Journal of Electromyography and Kinesiology. 15, 323-331, 2005

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PREFACE

In studies concerning spinal problems among helicopter pilots, pain from the neck region has until recently rarely been mentioned as a health problem. From the early 1960s up till now, numerous reports have appeared on helicopter pilots' lower-back pain as a consequence of flying (e.g. Borrillo, 1999; Bowden, 1987; Froom et al, 1987; Shanahan and Reading, 1984; Thomae et al, 1998). However, during the past two decades the prevalence of neck pain among helicopter pilots seems to have increased, and more and more studies have focused on the risk (Äng and Harms-Ringdahl, 2005; Bridger et al, 2002; Thomae et al, 1998). This increase in neck problems appears to have some connection with the increased use of night vision goggles, introduced among Swedish helicopter pilots in the 1980s.

In the early 1990s the Swedish Defense Research Agency acknowledged the increasing neck problems among Swedish military pilots (fixed-wing as well as helicopter, or rotary-wing). Cooperation was instituted with Professor Karin Harms-Ringdahl at the Karolinska Institutet and Colonel Jan Linder at the Aeromedical Section of the Armed Forces Headquarters. At first, the focus was primarily on fixed-wing pilots, but since the late 1990s the focus has shifted to rotary-wing pilots.

This collaboration has now resulted in this thesis concerning neck load among helicopter pilots and in an ongoing PhD-project by Björn Äng, concerning the effects on muscle performance of neck pain among helicopter pilots.

The present results will, it is hoped lead to better understanding of the effects of external neck loads on helicopter pilots. They should also serve as a reference in future studies of neck loads during flight missions.

Marcus Thuresson

TABLE OF CONTENTS

LIST OF ABBREVIATIONS

ANOVA	Analysis of variance
C	Cervical
CoM	Center of mass
CW	Counterweight
EMG	Electromyography
FFT	Fast Fourier transform
hCW	Helmet, night vision goggles and counterweight
HMD	Helmet-mounted display
hNVG	Helmet and night vision goggles
ICC	Intraclass correlation
IMDF	Initial median frequency
MU	Motor unit
MUAP	Motor unit action potential
MUR	Muscle strength utilization ratio
MVC	Maximum voluntary contraction
MVE	Maximum voluntary electrical activation
NVG	Night vision goggles
RMS	Root mean square
RVC	Reference voluntary contraction
S_w	Measurement error
T	Thoracic

1 Introduction

Helicopter pilots complain of pain originating from the neck region (Äng and Harms-Ringdahl, 2005; Bridger et al, 2002; Thomae et al, 1998). The causes and anatomical locations of the pain are basically unknown, but the ergonomic situation when flying a helicopter has been proposed as a causative factor (Thomae et al, 1998).

When flying the helicopter the pilots sit in a fixed posture for long periods in an asymmetric position due to the complex maneuvering of the helicopter. A helicopter is maneuvered partly with the feet, partly with the right hand, which operates the cyclic pitch stick between the pilot's legs, and partly with the left hand on the collective pitch lever to the left of the pilot. As a cause of the layout of the controls the pilot has to lean forward with the trunk slightly rotated to the left (Lopez-Lopez et al, 2001). In addition to the sitting posture, flight missions requires head movements with neck positioned away from neutral (Rostad et al, 2003a; Rostad et al, 2003b; Verona et al, 1986), causing additional load on the cervical spine. When using night vision goggles (NVG) the field of vision decreases thus increasing the range of head movement (Rostad et al, 2003a).

The helmet used is primarily for protection. Its weight depends on the level of protection required and the electronic equipment inside it (McEntire, 1998).
When flying, the helicopter pilots in the Swedish Armed Forces use a crash helmet model "Alpha" with a weight of about 1.5 kg, designed to withstand minor as well as major blows.
As well as protection, the helmet is used for attaching helmet-mounted displays (HMDs) such as NVG mounted to the front. NVGs take the small amount of ambient light (e.g. moonlight or starlight), and convert the light energy (photons) into electrical energy (electrons) (ITT Industries, 2005). In the image intensifier, photons enter a photocathode, which in turn releases electrons. These pass through a thin disk (a microchannel plate) containing over 6 million channels. As the electrons pass through the channels, they strike the channel walls, releasing more electrons. By the time the electrons leave the microchannel plate, they have been multiplied thousands of times. The multiplied electrons then are accelerated onto a phosphor screen. Electrons striking the screen cause it to glows in the same pattern as the light (photons) that originally entered the photocathode. The user then sees a brightened, intensified and somewhat greenish image (ITT Industries, 2005). This process requires both energy and space, and thus NVGs usually weight about 700 grams, not counting a battery pack mounted on the back of the helmet. The evolution of NVGs has resulted in better image quality, but the weight has remained about the same.

The main rotor blades above the helicopter cabin press air downwards to achieve lift. Each time a blade passes the cabin, it sustains a blow from the air pushed downwards, which shakes it. This causes a constant, low-level, mainly sinusoidal, vertical whole-body vibration inside the helicopter cockpit. The frequency of the vibration depends on the number of blades on the rotor and the speed of rotation.

The magnitude of the vibration depends on many variables, but in general, the fewer blades, the higher vibration magnitude. However, were there too many blades, the lift they should create would vanish. Most modern helicopter rotors have four blades.

1.1 Overall aim

The overall aim of this thesis was to quantify effects of external loads on helicopter pilots' necks, induced by e.g. neck and body positions, head-worn equipment and whole-body vibration, and the interactions among these, and to evaluate methods for measuring neck load.

1.1.1 Specific aims

1.1.1.1 Neck and body position

The effect of different neck and body positions was evaluated concerning muscle activity (studies I, II, III, and IV), mechanical load (study II) and seat-to-head vibration transmissibility (studies III and IV). The positions evaluated were: neutral (studies I-IV), trunk inclined 20° (study I), neck flexed 20° (studies I-III), and neck rotated 30° left and right, respectively (study I).

1.1.1.2 Head-worn equipment

The effect of different head-worn equipment was evaluated concerning muscle activity (study I-III), mechanical load (study II) and seat-to-head vibration transmissibility (study III).
The head-worn equipment evaluated was: helmet, helmet + NVG, and helmet + NVG + counterweight.

1.1.1.3 Whole-body vibration

The effect of whole-body vibration on muscle activity (studies III and IV) was evaluated, and on seat-to-head vibration transmissibility (study III and IV).
The vibration characteristics evaluated were: vibration frequency from 2.5 to 30 Hz (studies III and IV) and vibration magnitude at 0.5, 1 and 2 m/s^2 (study IV).

1.1.1.4 Reliability of fatigue protocol

The test-retest reliability of a protocol aimed to assess helicopter pilots' neck-muscle fatigue was evaluated concerning the slope and the normalized slope of the median frequency decline, the initial median frequency, and subjective ratings of neck-muscle fatigue (study V).

1.2 Theoretical framework

Physiotherapy can be defined as 'a health care profession concerned with human function and movement and maximizing potential: It uses physical approaches to promote, maintain and restore physical, psychological and social well-being, taking account of variations in health status; It is science-based, committed to extending, applying, evaluating and reviewing the evidence that underpins and informs its practice

and delivery, and the exercise of clinical judgment and informed interpretation is at its core.' (The Chartered Society of Physiotherapy, 2002).
With these definitions it is easy to see the interaction between physiotherapy and ergonomics, where the physiotherapist functions as an expert in relating environmental load, ergonomics, body function and structure pathology. Ergonomics has been described as comprising three elements: craft, science and engineering (Long and Dowell, 1996). It aims to implement and evaluate (craft), to explain and predict (science), and to design for improved performance (engineering).
Chapinis (1996) defines ergonomics as 'a multi-disciplinary field, with psychology, anthropometrics, applied physiology, environmental medicine, engineering, statistics, operations research and industrial design all contributing. Ergonomics is both a discipline and a profession; the field of study is the theory and practice of understanding people and their characteristics in relation to design'. In addition to field measurements, laboratory research also has a place in ergonomics; it is possible to improve interactions for people at work, at home and at leisure through well planned and interpreted experiments (Wilson, 2000). One major advantage of laboratory studies is that they permit changes of risk factors one at a time, at the same time controlling for confounding factors. In aviation medicine this is obvious since it is expensive and complicated to perform evaluations during flight: it is important to perform well-planned laboratory studies first.

The role of ergonomists is two-fold (Wilson, 2000). First, they need to fundamentally understand purposive interactions between people and artifacts and especially to consider the capabilities, needs, desires and limitations of people in such interactions. In this role ergonomists are scientists, embracing qualitative enquiry in the field as much as controlled laboratory experimentation. Secondly, ergonomists contribute to the design of interacting systems, maximizing capabilities, minimizing limitations, and trying to satisfy human needs and desires. Here ergonomists are craftspeople, using judgment, vision, experience and even trial and error to develop and test concepts and prototypes (Wilson, 2000).

A physiotherapist's occupational role most often concerns musculoskeletal ergonomics. Here, the focus of analysis should be the interaction between individuals, the task and the workplace (Hägg et al, 2000). Prevention of musculoskeletal disorders is a cornerstone of ergonomics, and even if the mechanical load on the human body in working life is not the only causative factor, it is likely to constitute a major part of it (Hägg et al, 2000). Therefore estimation of physical exposure, and its physiological consequences, are essential activities in physiotherapeutic occupational work.

Mechanical loads applied on joints, muscles or tendinous structures may cause pain, of great importance in ergonomics (Harms-Ringdahl, 1986). In this context, analysis of the genesis of load-elicited pain is of central importance in physiotherapy. A few authors have presented different, but similar, models concerning the relationship between mechanical exposure and health effects. Harms-Ringdahl showed how induced load moment counteracted by neck structures can cause load-elicited pain (Harms-Ringdahl, 1986). A more general model was presented by Westgaard and Winkel (1996) indicating the relationship between physical work load and musculoskeletal health effects. In those authors' model intermediate stages in the relationship were also presented; such as biomechanical forces generated to meet the demands and short-term physiological and psychological responses. Effect modifiers influence the link between the different elements in the model, related both to the environment and to the individual.

1.3 Biomechanics

Biomechanics can be defined as 'the application of the principles of mechanics to the study of biological systems' (Enoka, 1994). Biomechanics uses laws of physics and engineering concepts to describe the motion of the various body segments and the forces acting on these during normal daily activities (Frankel and Nordin, 1980). In ergonomics, external load is commonly caused by forces acting on body parts or equipment, and internal load is caused by muscles or other soft tissues.
There is no simple way to measure the total mechanical load applied to the human body (Kadefors, 1978), but several measures may give estimations of the load. Such measures can be objective (e.g. registration of muscle activity, biomechanical calculations or video recordings) or subjective (e.g. interviews or questionnaires).

1.3.1 Cervical-spine biomechanics and anatomy

The human cervical spine can has two structural components - hard and soft tissues. The hard tissues include the vertebrae and intervertebral disks. Their function is primarily load-bearing: they resist compressive forces. The hard tissues of the head and cervical spine can be divided into three distinctive biomechanical components.
The head is attached to the upper cervical spine at the first cervical vertebrae (C_1) via the occipital bone (C_0). The second component, the upper cervical spine, consists of the atlas and axis (C_1 and C_2). C_1 has no vertebral body and there is no intervertebral disk between it and C_2. The C_0-C_1 joint allows primarily flexion-extension motion and the C_1-C_2 joint primarily axial rotation. Functionally, C_0-C_1-C_2 acts as a ball joint. The third component is remaining, typical cervical spine (C_3-C_7), which function more similarly to the rest of the spine.

The soft tissues – the muscles and ligaments - act mainly to stabilize the neck and to provide for head movement. The combined motion of all segments of the cervical spine produces a large range of motion – about 140 degrees of flexion/extension, about 180 degrees of axial rotation and approximately 90 degrees of lateral flexion (Dvorak et al, 1992). To accomplish head movement and stabilization, over 20 pairs of muscles reach the head (Kamibayashi and Richmond, 1998). The dorsal neck muscles are positioned in four different layers (Mayoux-Benhamou et al, 1997). The innermost layer is generally primary stabilizers and the outer three are responsible for movement.
To move the head, several muscles counteract, and their function differs depending on neck position. However, when performing isometric contractions with the head in a

neutral position, semispinalis capitis is activated during neck extension (Keshner et al, 1989; Queisser et al, 1994; Takebe et al, 1974), while splenius capitis is activated during ipsilateral rotation (Mayoux-Benhamou et al, 1997; Takebe et al, 1974) and lateral bending (Keshner et al, 1989; Mayoux-Benhamou et al, 1997), in addition to neck extension (Mayoux-Benhamou et al, 1997; Takebe et al, 1974),. Sternocleidomastoid is the primary neck flexor (Keshner et al, 1989; Mayoux-Benhamou et al, 1997), and is also activated during contralateral rotation and lateral bending. Trapezius has little or no effect on head movement (Keshner et al, 1989).

1.3.2 Electromyography (EMG)

Due to the neuromuscular function of a contracting muscle, leading to an electric current propagating along the muscle fibers, muscle activity can be recorded using electromyography (EMG). Assessing activity in the muscles through electromyography provides insight into patterns of activation or indication of intrinsic tension developed in the muscles. This may be of interest in itself, because sustained muscle activity might cause ischemic muscular pain (Sommerich et al, 2000).

Either intra-muscular electrodes or surface electrodes can be used to detect the action potential's electrical characteristics over the muscle fibers. Intra-muscular electrodes selectively detect activity from within muscles, whereas surface electrodes pick up activity from a more widespread area (Mayoux-Benhamou et al, 1997). In the neck region only a few superficial muscles are selectively detectable using surface EMG. These include the sternocleidomastoid and upper trapezius muscles. Registration of other superficial muscles, such as the splenius capitis and semispinalis capitis is likely to contain muscle activity from adjacent muscles as well (Benhamou et al, 1995), making statements about the activity of those specific muscles uncertain. For these muscles, placement of the electrodes can be location-specific rather than muscle-specific.

In biomechanics, three applications dominate the use of the surface EMG signal: its use as an indicator of the initiation of muscle activation, its relationship to the force produced by a muscle, *i.e.* amplitude properties, and its use as an index of fatigue processes occurring within a muscle, *i.e.* its spectral properties (De Luca, 1997). In this thesis, the latter two are of major interest.

1.3.2.1 Amplitude properties

Since the EMG signal can be regarded as stochastic, the amplitude is usually expressed as a root mean square average (RMS), an average rectified value, or an integrated rectified value (Basmajian and De Luca, 1985; Hägg, 1991). These amplitude estimations show similar response to force alternations, but the RMS value is recommended above the others (Basmajian and De Luca, 1985). As the number of motor units (MU) increases and/or the firing frequency of the activated MUs increases, the RMS value of the EMG signal increase. As MU activation reflects the force required, the RMS value can be used for estimating external mechanical load. However, this relationship depends on many physiological, anatomical, and technical factors (De Luca, 1997); the force/EMG relationship differs substantially between muscles and is rarely linear. To adjust for variation due to differences in electrode spacing, anatomical factors, and variations in electrode placement in multi-day experiments, and to allow comparison between different muscles and subjects, the RMS value should be normalized (Sommerich et al, 2000). Normalization is also a prerequisite for including an EMG study in a meta-analysis of occupational exposure and its musculoskeletal effects (Mathiassen et al, 1995). The RMS value can be related either to the attempted maximum muscular activity (expressed as % MVE) or to a contraction at a specified sub-maximal level (expressed as % RVE) (Mathiassen et al, 1995). In this thesis, muscle activity is related to maximum effort and is thus presented as % MVE.

1.3.2.2 Spectral properties

EMG generally reflects a large number of motor-unit-action-potential (MUAP) trains with (theoretically) independent firing frequencies (Hägg, 1991). The EMG signal can be split up into its frequency components using a Fast Fourier Transform (FFT) and a power spectrum can then be presented (Basmajian and De Luca, 1985) (Figure 1). To describe the frequency content, as with any independent sample, a central measure is often employed, such as the mean or the median. In the literature both measures are used, and both have their supporters; but due to its lower sensitivity to noise and signal aliasing, the median frequency has been suggested to be the best measure (De Luca, 1984). From a statistical point of view, due to the asymmetric shape of the power spectrum, the median frequency would be advocated. The shape of the power spectrum depends on several factors, such as the power spectrum of each MUAP, the firing rate for each MUAP train, and the filtering properties of the electrode configuration (Basmajian and De Luca, 1985).

Some extrinsic factors can also contribute to differences in the appearance of the power spectrum. These include differing force levels, electrode distances, and temperatures. Sustained contractions alter the power spectrum into lower frequencies. Thus the evaluation of the median frequency change over time can be used as an index of muscle response to sustained external load. Typically, median frequency decreases linearly and a regression line can be approximated. Further, the slope of the regression line can be used as an estimation of muscle fatigue, a steeper slope indicating a more fatigued muscle.

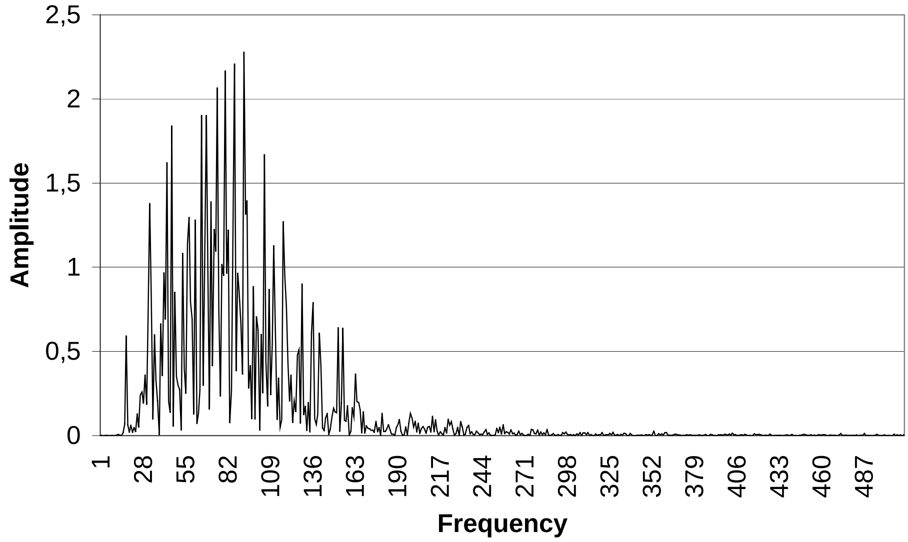

Figure 1. An example of a power density spectrum of a contracting muscle.

1.3.3 Biomechanical models

The mechanical load on the cervical spine can be estimated using biomechanical models. Biomechanical models can broadly be divided into four groups (Panjabi, 1998): *i*) physical models, made of non-anatomic material; *ii*) in vitro models consisting of a cadaveric spine specimen; *iii*) in vivo animal models; and *iv*) computer models developed from mathematical equations. A computerized model can in its simplest form be a static two-dimensional model, but it can also be a dynamic, three-dimensional multi-segment model (Kumaresan et al, 1999). More sophisticated models have also been presented for estimating the internal load induced by the neck muscles (Vasavada et al, 1998). A computerized biomechanical model might be complicated and expensive to produce, but it has the advantage of not exposing subjects to any potential harmful mechanical load. Once produced, it can be used to evaluate many different experimental settings without too much effort. In the present work, a simple computerized model was used.

1.3.4 Subjective ratings

Mechanical load can also be estimated more subjectively using a measurement scale or questionnaire. Advantages are that the method is cheap and easy to administrate, and different types of biomechanical load can be estimated. There are also drawbacks. Since the response is based on subjective values among the individuals evaluated, it is sensitive to the responder’s ability or wish to respond, which could jeopardize the validity of the method as a measure for evaluating exposure to mechanical load (Wiktorin, 1995).

1.4 Literature review

This review does not seek to cover the whole area, but rather to give additional information concerning the biomechanical and physiological effects of the risk factors addressed.

1.4.1 Physiological effects of flying a helicopter

Muscle activity in the back muscles during flight has been recorded in a few studies with somewhat varying results. Lopes-Lopes (2001) found an increase in muscle activity in the back muscles when flying, and especially in the right side muscles. On the other hand, de Oliveria et al (2001; 2004), found no effect of flying on EMG activity, neither as an effect of sitting position or of vibration.
Hewson et al (2000a; 2000b) evaluated forces applied on the sticks and corresponding muscle activity in the arms and legs during certain maneuvers. The authors concluded that during routine maneuvers only small forces are required on the sticks, and muscle activity is low (Hewson et al, 2000b). However, during emergency maneuvers the maximum required forces increase most on the cyclic stick, sometimes tenfold (Hewson et al, 2000a). For the collective stick and pedals, there were higher forces than those in routine maneuvers. The corresponding mean muscle activity also increased from about 5-10 % MVE during routine maneuvers to 35-45 % MVE during some emergency landings.

A few reports have focused on head motion during flight (Rostad et al, 2003a; Rostad et al, 2003b; Verona et al, 1986). Verona et al reported that for 95 % of the time the pilots looked between 14° flexion and 14° extension. The range was from 65° flexion to 30° extension. For the rotation analysis the range was larger (+-90°) but for most of the time the head was held in a relatively neutral position (within +- 20° rotation). Rostad (2003a; 2003b) evaluated head movements when using different visual aids such as night vision goggles. During this test the pilots flew a familiar route with no potential threats, as were the case in Verona's et al. study. As a consequence the range when flying without visual aids (as in Verona's study) was smaller. Of interest for the present work is the comparison between flying with NVG and flying with helmet only. The analysis of flexion/extension revealed only minor differences between the two setups (Rostad et al, 2003b), but the standard deviation and range when flying with NVG was larger (Rostad et al, 2003a), which is likely due to the decreased field of view when using NVG.

1.4.2 Effects of neck and body position

1.4.2.1 Biomechanical effects

Sitting in a normal, upright position with the head in a neutral position causes generally low load on the cervical spine (Harms-Ringdahl, 1986). The load moment is balanced by muscle forces and tension of the passive structures. The more the head departs from neutral, the more the load increases; measured both as induced load and as muscle activity (Harms-Ringdahl et al, 1999). However, when the head is held in a maximum flexed position, the muscle activity in similar to that in neutral position, even though the induced load above the C_7-T_1 motion segment is increased 3-4 times (Harms-Ringdahl et al, 1986). A comparison of different trunk and neck positions shows that the position with lowest neck muscle activity is obtained when the trunk is slightly backward-inclined and the head held vertical (Schüldt et al, 1986). Finsen (1999) evaluated neck load and EMG activity in moderately and highly flexed neck positions. The induced load in the lower cervical spine increased at the highly flexed position, but for the upper cervical spine load there was no difference between positions. In addition, splenius muscle activity decreased at the highly flexed position (though not significantly). These results indicate that, in extreme positions the induced load is sustained mainly by the passive structures in the neck and not by the neck muscles.

1.4.2.2 Physiological

Prolonged sitting with the neck in extreme positions may cause neck pain (Harms-Ringdahl and Ekholm, 1986). However, the literature concerning long-term effects of this issue is far from united. In a review, Ariëns et al (2000) found some evidence for a positive relationship between neck pain and e.g. neck flexion and prolonged sitting, based on four 'low-quality' studies. They concluded that there was insufficient number of high quality studies to base the conclusions on. For the relation between neck rotation and neck pain they found only two 'low-quality' studies, of which one found a positive relationship. A review by NIOSH (1997) reported strong evidence for a causal relationship between general posture and musculoskeletal disorders in the neck and shoulder. Of 31 reviewed articles, 27 reported statistically significant positive association.

1.4.3 Effects of head-worn equipment

1.4.3.1 Biomechanical effects

To meet the need for extensive protection, a helicopter helmet normally weighs about 1.5 kg. In addition to the relatively high weight, the center of mass usually lies above and in front of the head's center of mass. With HMDs the weight increases and the center of mass usually shifts even further forward. A few studies report the biomechanical effects of bulky head-worn equipment, measured as muscle activity, muscle fatigue, and induced mechanical load. Philips and Petrofsky evaluated neck loads induced by head-worn equipment (Petrofsky and Phillips, 1982; Phillips and Petrofsky, 1983a; Phillips and Petrofsky, 1983b; Phillips and Petrofsky, 1984; Phillips and Petrofsky, 1986). The found that the muscle endurance decreased when extensive head-worn equipment was used (Phillips and Petrofsky, 1983a). The EMG parameters evaluated also showed signs of a fatiguing reaction (Phillips and Petrofsky, 1983b).

Mechanical load at different neck angles and under different gravitational force levels (G_z) when using NVGs has been evaluated (Harms-Ringdahl et al, 1991). The NVG increased the mechanical load but a counterweight reduced the load. However, the more the neck was flexed the more the load-reducing effect of the counterweight decreased, and eventually the counterweight induced a flexing moment (Harms-Ringdahl et al, 1991).

Butler (1992) evaluated neck muscle response and seat-to-head transmissibility (see 1.5.4.1) due to whole-body vibration when using a variety of helmet weights and load moments, This author suggested a limit of flexing load moment of 8.27 Nm about the atlanto-occipital complex. Further he suggested that the load moment should not be negative i.e. the center of mass (CoM) of the head-worn equipment should not be located behind that of the head.

1.4.3.2 Physiological effects

Little is known about the physiological effects of wearing increased weight on the head. Documentation concerning neck injuries during helicopter crashes shows that increased helmet weight gives higher risk (McEntire, 1998), as does the use of night vision goggles (Shannon and Mason, 1997). In an ongoing study (Äng and Harms-Ringdahl, 2005) concerning helicopter pilots' neck pain, a Cox regression analysis revealed frequent use of NVG to be a risk factor for neck pain (relative risk of 1.6). Of the pilots using NVG, 74 % reported neck pain within the previous three months as compared to 41 % of those who had never or rarely used NVG. Apart from anecdotal reports of neck pain and strain as a result of prolonged use of increased head loading, few studies have scientifically reported such effects. Jager et al (1997) reported spinal changes among African women after carrying substantial load on their heads.

1.4.4 Effects of whole-body vibration

1.4.4.1 Biomechanical effects

The biomechanical effects of whole-body vibration have long received considerable attention. Evaluation of the transmissibility of vibration through the human body reflects the various biodynamic responses of the body, particularly those between the point at which the vibration enters, and that where it is measured (Paddan and Griffin, 1998). Typically, the characteristics of the vibration are divided into two components – acceleration magnitude and frequency. Assuming that the resonances in e.g. seat-to-head transmissibility indicate frequencies at which injury, discomfort or interference with activities are most likely, major concerns have been the effects of frequency and the human body's resonance frequencies (Paddan and Griffin, 1998). Driving point impedance, apparent mass, seat-to-head transmissibility, and abdominal pressure response have all been evaluated with respect to whole-body vibration frequency at vibration magnitudes ranging from about 0.1 to 5 m/s^2. No matter what evaluation method used, these studies all report a principal resonance frequency of the seated human body at about 4-5 Hz.

In the evaluations of the effects on humans of different magnitudes of whole-body vibration, no clear effect has been found. Holmlund et al (2000) found an decrease of mechanical impedance due to increase of acceleration magnitude in the frequency range 5-20 Hz, and also that the frequency of the resonance peak decreased with the vibration magnitude, also reported by Holmlund and Lundström (2001). Similar results were also found by Mansfield and Griffin (2000) where the resonance peak of the apparent mass decreased as the vibration magnitude increased. In addition, the level of the peaks increased slightly. In another study, seat-to-head transmissibility decreased as

acceleration magnitude increased (Griffin, 1975) at frequencies between 7 and about 30 Hz.
Muscle activity in the neck muscles as a response to whole-body vibration has also been reported in a few studies. Cheng (1996) evaluated neck muscle activity during exposure to random vibration during 4 hours. He found among that there was a distinct effect after 130 minutes, where the muscles fatigued. Butler (1992) evaluated muscle activity as a response to vertical sinusoidal whole-body vibration and reported changes in burst sizes and timing in relation to the vibration frequency, where the largest burst was achieved near the resonance frequency.

1.4.4.2 Physiological effects

Whole-body vibration can be a risk factor for neck pain, though this is not undisputed (Wikström et al, 1994). A higher vibration magnitude reportedly causes more discomfort in the neck and shoulder, especially in combination with non-neutral neck postures (Wikström, 1993). Among drivers of all-terrain vehicles the prevalence of neck pain is about twice that in a control group, and it has been suggested that this is caused by shock and vibration, static overload of the neck muscles and strenuous and extreme neck postures (Rehn et al, 2002). This was partly verified in a subsequent study where the characteristics of neck pain differed from those of the controls (Rehn et al, 2004). One radiographic study among helicopter pilots has indicated an increase in cervical osteoarthritic changes as compared to other flying groups and a control group (Aydog et al, 2004) It was hypothesized that this increase was caused by whole-body vibration. In the NIOSH (1997) review cited above, too few studies were found to permit any statement about the causal relationship between whole-body vibration and musculoskeletal disorders in the neck and shoulder.

1.4.5 Neck muscle fatigue

Evaluation of selective muscle fatigue in the cervical muscles using electromyography is complicated since more than 20 pairs of muscles reach the head, thus affecting its movement and stabilization (Kamibayashi and Richmond, 1998). Reliable techniques are urgently needed since estimation of muscle fatigue can be used both in clinical evaluation of neck pain and when evaluating physical demands at a workplace.

More than 20 years ago, Phillips and Petrofsky (1983b) concluded that surface EMG of neck muscles could be used as a non-invasive, objective and quantitative index of neck muscle fatigue. They reported a greater decline in the median frequency of the cervical muscles during static contractions after exposure to a heavier combination of head-worn equipment than after exposure to a lighter one. However, few studies have since used surface EMG to evaluate neck muscle fatigue. Gogia and Sabbahi (1990; 1991; 1994) found that fatigue levels differed between fatigue testing in supine position and in erect position (Gogia and Sabbahi, 1991). Patients suffering from osteoarthritis in the cervical spine were more easily fatigued in both the anterior neck muscles and the posterior neck muscles than healthy controls under both moderate and high loads (50 and 80% MVC) (Gogia and Sabbahi, 1994). Similar results were found by Falla et al (2003) where

chronic neck pain patients were more easily fatigued in the cervical flexors at low and moderate loads (25 and 50% MVC) than healthy controls were. Somewhat contradictory, Äng et al (Äng et al, 2005) reported that helicopter pilots with frequent episodes of neck pain had less steep slopes of median neck flexor frequency, measured when they were pain-free, possibly indicating changes in muscle morphology induced by the loads sustained when flying helicopters.
There are indications that pain and specific training can influence neck muscle EMG parameters. An eight-week resistance training program for the lateral neck muscles decreased the slope of the mean power frequency and increased endurance time and maximum isometric torque (Portero et al, 2001).

Besides evaluation of the frequency spectrum as an index of muscle fatigue, other techniques have also been used. Oksa et al (1999) created a fatigue index based on maximal isometric strength and peak strain during flight among a group of fighter pilots exposed to high G forces, comparing before and after flight. The most fatigued muscles after repeated aerial combat were the dorsal neck muscles and the sternocleidomastoid (other muscles compared were the abdomen and back muscles).
Another way of measuring neck muscle fatigue is to evaluate the endurance time to exhaustion. This has been used in the aviation field by e.g. Alricsson and colleagues (2001) evaluating jet pilots' neck muscle endurance. The pilots had poorer endurance than a group of conscripts without flying duties.

1.4.6 Reliability

Reliability can basically be defined as how far a measurement instrument gives the same results with repeated measurement, assuming that whatever is being measured does not change (Neuman, 1997). Other sub-definitions of reliability are: agreement, consistency, conformity, reproducibility and repeatability (Müller and Büttner, 1994). Agreement is often used when two (or more) instruments are compared or the agreement between two (or more) raters using the same instrument (inter rater reliability) (Bartko, 1994) is compared. Conformity is defined as a measurement's agreement with a 'standard' reference (Müller and Büttner, 1994). Consistency, reproducibility and repeatability are often used when the same examiner is assessing one measurement twice or more (also known as intrarater reliability or 'a test-retest design') (Müller and Büttner, 1994).

The reliability of using EMG for recording muscle activity level and fatigue has been tested for several human muscles with varying results. For the neck muscles, there are few reliability studies. Veiersted (1996) reported acceptable reliability for static trapezius muscle activity during repetitive light manual work among female workers. The lack of studies of the reliability of neck muscle EMG activity levels has also been addressed elsewhere (Sommerich et al, 2000). The reliability of surface EMG evaluations of work tasks in other muscles has been studied. For example, Danneels et al (2001) evaluated back muscle activity and fatigue during 22 exercises and found that the only reliable measure was the evaluation of intraoperator average EMG.

Otherwise, the main focus has concerned the spectral properties of the EMG signal. Strimpakos et al (2005) reported ICC values between 0.28 and 0.61 for the normalized slope of the median frequency of four cervical muscles during 30s contractions at 60 % MVC. In addition those authors reported high S_w values. The initial median frequency showed better reliability parameters with ICC between 0.64 and 0.81. Similarly, Gogia et al (1991) reported high ICC values (intraclass correlation) for the initial median frequency for a method of measuring neck muscle fatigue during brief exposure. Those authors did not present a measurement error to quantify the within-subject variation. Falla et al (2002b) reported poor ICC values for the slope of the mean frequency of the sternocleidomastoid muscle.
Since the reliability values are based on within- and between-subject variance, the results apply only to that specified population. Hence it is important that the group tested in a reliability study is equivalent to the target population for which the test is designed.

1.5 Statistics

1.5.1 Statistics in repeated-measures studies

Repeated-measures is the term for data in which the response of each individual is observed on several occasions or under varying conditions. There are both advantages and disadvantages when taking several measurements on each individual. In repeated-measures design one can obtain individual patterns of change, which is not possible when observing different individuals at each time point. This design also minimizes the number of individuals. The same individual can also be measured under both control and experimental conditions so that one individual can be his own control, which minimizes individual variation. One problem with repeated measures design is that the measurements on the same individual are not independent, and this has to be considered. Further, the measurements should not be affected by the previous measurement occasion.

One method that takes advantage of the repeated-measures design and allows for adjustment due to the dependence between measurement occasions is the *mixed models design*. This gives a few positive effects not present in ordinary analysis of variance (ANOVA). First, the items within the random effects may have different degrees of inter-correlation, which can be modeled. Secondly, the mixed models design admits missing data, and still uses the information that can be found.
The *mixed linear models* are based on assumptions about the distributions of the study population and of the residuals. As for *linear models without random effects* it is important to check for these assumptions. In addition to the assumptions in linear models the mixed model also make assumptions concerning random effects, which should be approximately normally distributed. This suggested the following twofold approach: *i*) embed the model control in linear models without random effects by considering a model in which all random effects are taken as fixed; and *ii*) do a simple and rough control of assumptions as to the normality of the random effects.

The "classical" assumptions for linear normal models (without random effects) are that: *i*) the model structure should capture the systematic effects in the data; *ii*) normality of residuals; *iii*) variance homogeneity of residuals; and *iv*) independence of residuals Apart from the formal assumptions it is important to take account of any outliers and other influential observations

1.5.2 Statistics in reliability studies

Several statistical methods have been used in reliability studies depending on the focus of interest. Bland and Altman recommend that the differences or the standard deviations of the measurements should be plotted against the means of the measurements to see if the differences or the standard deviations are related to the means (Bland and Altman, 1996). If they are, the results cannot be used in their original form and should be logarithmically transformed before statistical analysis.
After the control of the results with the Bland-Altman plots there are several statistical lines that can be followed. At present two measurements are recommended in reliability studies, which considered complementary (Keating and Matyas, 1998). The intraclass correlation coefficient (ICC), presents a relative outcome of reliability, while the measurement error (see below), which is the within-subject standard deviation (S_w), presents an absolute measure of reliability.

There are several variations of the intraclass correlation coefficient, but basically it is the ratio of the variance of interest to the variance of interest plus the error (Shrout and Fleiss, 1979). The differences between the variations of ICC are lie chiefly in how they treat the error term. A high value of ICC (close to 1) indicates good reliability, i.e. low within-subject variance in relation to between-subject variance. However, since the ICC is variance-dependent, the correlation can be very low even though there is an almost perfect match if the variation between subjects is very small (Bartko, 1994).
In addition to the ICC it is also important to present an absolute measure of reliability. The measurement error is the within-subject standard deviation and is a measure of how much the results are expected to vary between the test occasions due to random errors (Bland and Altman, 1996). In contrast to the ICC value, the S_w is an absolute measure of the same unit as the measured one and is hence easier to implement clinically.

2 Subjects and methods

2.1 Subjects

In all the investigations reported in this thesis Swedish military helicopter pilots participated. Their personal characteristics are presented in Table I. Thirty-nine pilots were included, of whom eight participated in two of the studies and two in three (studies I and II are counted together here). For study I the pilots were recruited consecutively at the Aeromedical Section of the Armed Forces Headquarters and for studies III-V the pilots were recruited among pilots working in the Swedish Armed Forces Headquarters, the Swedish National Defense College, and from two helicopter squadrons. During all of the tests the pilots were pain-free in the neck region and only a few had suffered from neck problems the month preceding the tests.

Table I. Characteristics of the pilots participating in the investigation. Age, height and weight are presented as mean and standard deviation and flying hours as median and range.

Study	n	Age		Height		Weight		Total flying hrs		Flying hrs 3 months prior to study	
I & II	14	40.1	(10.4)	1.80	(0.04)	83.3	(7.2)	Not asked		60	(15-105)*
III	17	40.8	(5.8)	1.82	(0.06)	87.1	(10.8)	1700	(450-4110)	0	(0-50)
IV	10	41.1	(8.2)	1.84	(0.04)	84.3	(8.8)	1400	(550-4000)	7.5	(0-45)
V	10	40.1	(6.0)	1.81	(0.05)	86.0	(7.8)	1600	(500-3000)	0	(0-80)

* Derived from a question about monthly flying hours.

2.2 Test configurations

2.2.1 Neck and body positions

In study I nine different neck and body positions were evaluated (Figure 2). In studies II and III two neck positions were evaluated (neutral and 20° flexed) and in IV and V the neck was held in a neutral position. In studies I, II and V the pilots sat upright during the tests and in studies III and IV they sat in a helicopter mock-up, in a position similar to that when flying.

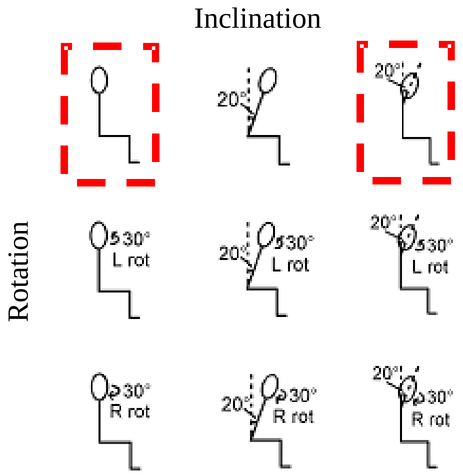

Figure 2. Test positions in study I evaluated concerning muscle activity. Positions shown in boxes were evaluated in study II concerning mechanical load and in study III concerning vibration transmissibility in addition to muscle activity recordings.

2.2.2 Head-worn equipment

In studies I-IV an Alpha military aircrew helmet (Helmet Integrated Systems Ltd, Stranraer, Scotland) was used. In studies I and II night

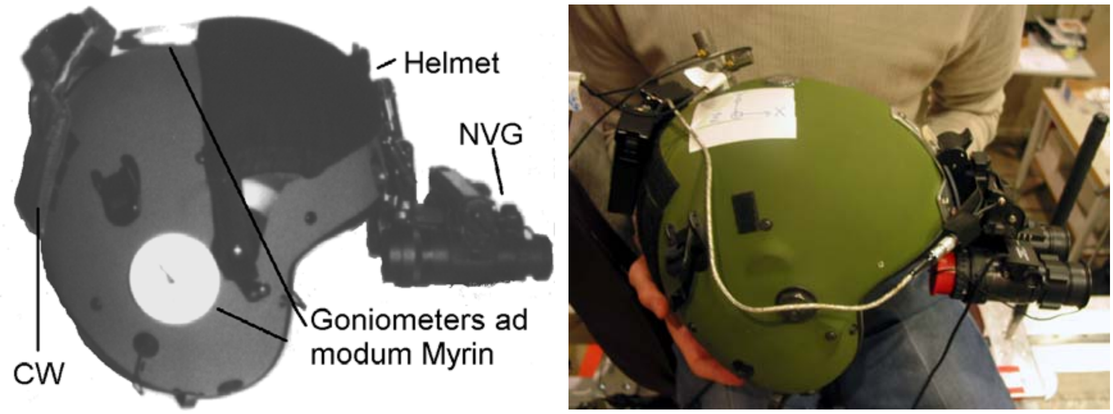

Figure 3. The helmet used in studies I and II (left) and in studies III and IV (right). On the helmet to the left the batteries were in the NVG and on the helmet to the right the battery pack was mounted on the back of the helmet, connected with a cable. On the right helmet the head-accelerometers are also mounted.

vision goggles (NVG) weighing 755 g and a counterweight (CW) weighing 325 g were added (Figure 3).
In study III night vision goggles (AN/AVS-6, ITT industries, White Plains, NY, USA) weighing 790 g and were attached in front of the helmet with a battery pack (240 g) attached to the back of the helmet with Velcro. Finally a counterweight (CW) (a bag containing lead plates) weighing 150 g, Velcro-ed to the back of the battery pack was used (Figure 3).

2.2.3 Whole-body vibration

In studies III and IV the pilots were exposed to sinusoidal vibration swept from 2.5 Hz up to 30 Hz at a rate of 1.43 octave/min, and then directly back again, from 30 Hz down

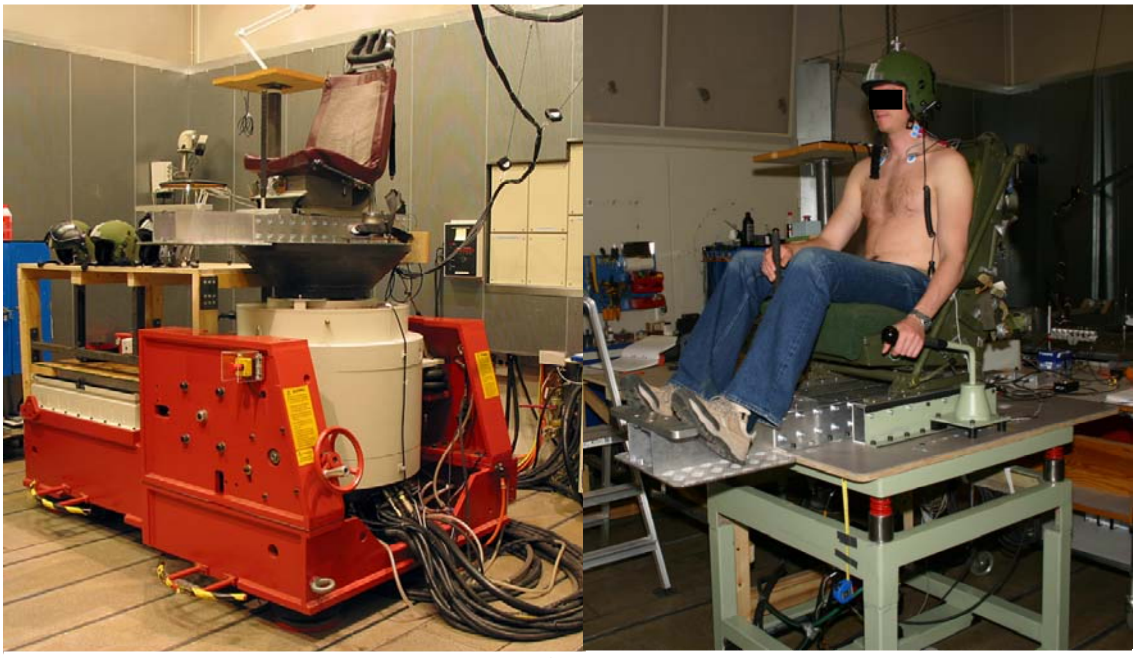

Figure 4. The vibration set-ups in study III (left) and study IV (right). In study III an electrodynamic shaker was used with a helicopter seat from a Swedish Army Helicopter 9 - MBB BO 105 CB-3 mounted on top. In study IV a servo-hydraulic shaker was used with a helicopter seat from a Swedish Army Helicopter 3 – Agusta-Bell 204 – mounted on top.

to 2.5 Hz. In study III the excitation magnitude was 1 m/s^2, and in study IV the excitation magnitudes were 0.5, 1 and 2 m/s^2. In study III the vibration was induced by an electrodynamic shaker and in study IV by a servo-hydraulic shaker (Hydraulic Actuator MTS 242.03) (Figure 4). On top of the shaker a helicopter mock-up was constructed, complete with seat (in study III from helicopter 9 and in study IV from helicopter 3), and with sticks and pedals positioned as in the helicopters.

2.3 Evaluation methods

2.3.1 Electromyography

EMG activity was recorded from up to four locations in the cervical region using disposable pre-gelled surface disc electrodes (Blue Sensor M-00-S (studies I and II) and N-00-S (studies III-V), Ag/AgCl, Medicotest A/S, Denmark) (Figure 5): *i*) upper posterior neck muscles (upper neck) with the cranial electrode at the level of vertebra C_2 between the uppermost parts of trapezius and the sternocleidomastoid, i.e. over the splenius capitis, and the other electrode 20 mm caudally (Schüldt, 1988) (studies I-V); *ii*) lower posterior neck muscles (lower neck) at the level C_7-T_1 with the cranial electrode 30 mm lateral to the C_7, i.e. over the erector spinae and the other electrode 20 mm caudally (Schüldt, 1988) (studies I-V); *iii*) sternocleidomastoid muscle, between the caudal tendon and the middle part (Falla et al, 2002a) with an inter-electrode distance of about two cm (studies III-V), and *iv*) upper trapezius muscle, at the antero-lateral margin, midway between acromion and occiput (Schüldt, 1988) with an inter-electrode distance of about two cm (studies I, III and IV), all bilaterally, except in study II where only left side was analyzed.

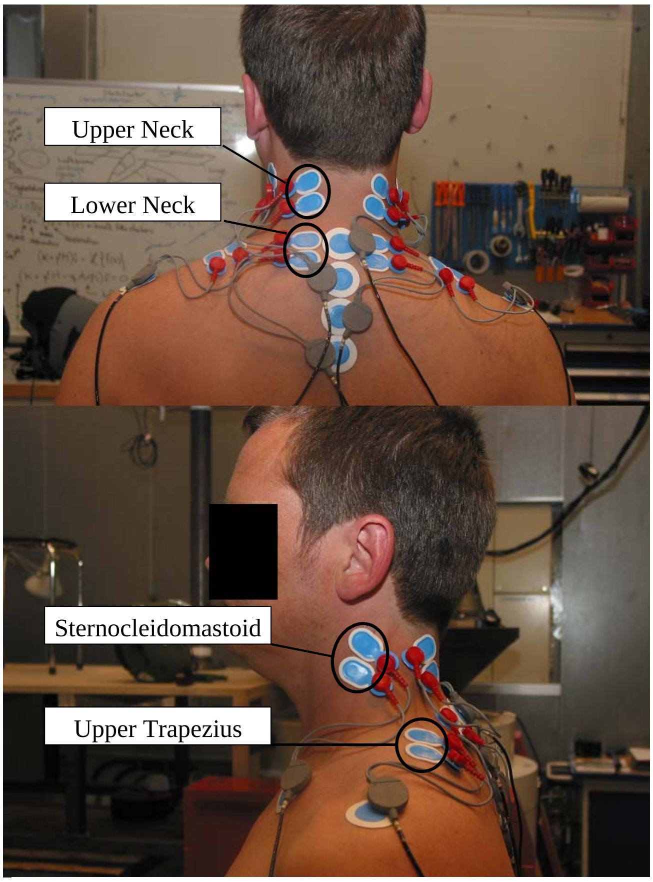

Figure 5. Electrode placements. Upper and lower neck were evaluated in all studies, trapezius in studies I, III and IV, and sternocleidomastoid in studies I and III-V. The gray buttons on the picture are ground electrodes with built-in pre-amplifiers.

Electrode attachment and normalization procedures followed recommendations (Hermens et al, 2000; Sommerich et al, 2000). The signals were amplified 1000 times in built-in pre-amplifiers in the EMG cables, band-pass-filtered 20-500 Hz, A/D converted and sampled at 1000Hz (Muscle tester ME3000P8, Mega Win 2.0, Mega Electronics Ltd, Kuopio, Finland).

2.3.2 Fatigue testing and EMG normalization procedures

The maximum isometric efforts (studies I, II and V) and the fatiguing contractions (study V) for the upper and lower neck and the sternocleidomastoid were performed sitting in a DBC140 equipment (David Back Clinic International, Vantaa, Finland) with the neck in a neutral position. For both neck extension and flexion the seat height was adjusted so that the bilateral motion axis of C_7-T_1 was in the same altitude as the motion axis of the DBC140. The chest-pad belonging to the DBC140 was firmly adjusted against the chest (extension) and back (flexion), respectively. In extension the subjects sat with the arms alongside the body, the hip as straight as possible and the knees flexed so that only the tip of the toe touched the floor (Äng et al, 2005). In flexion the hip and knee were flexed about 90°, the arms rested on the thighs and the neck was held slightly flexed.

In studies III and IV the maximum isometric efforts were performed sitting on a stool with a firm foam pillow between the wall and the chest (extension) or back (flexion), respectively. To monitor the applied force an S-beam load cell (Pt4000, Macmesin, UK) attached to a dynamometer (Macmesin AFG, UK) was secured to the wall and fastened to a horizontal sling around the subjects' head. In all other aspects the normalization position was the same as in studies I, II and V.
The MVC for upper trapezius (studies I – IV) was performed with 90° abduction of the arm in the scapular plane and elevation of the shoulder, with maximum isometric resistance applied manually to the shoulder and the upper arm proximal to the elbow joint (Schüldt and Harms-Ringdahl, 1988).

2.3.2.1 EMG analyses

<u>Studies I and II.</u> The raw EMG signals were rectified and averaged over the two middle seconds of each test position. Average muscle activity for each position was normalized against the MVE and is presented as a percentage of the MVE (% MVE) (in studies I and II presented as % RVC but performed as described). Before normalization calculations were performed, the lowest recorded values were subtracted from each test value, and thus the reported value is activity above baseline, including noise.
<u>Studies III and IV</u>. First a program was developed to calculate the root mean square (RMS) value of the EMG and accelerometer signals for one-second intervals. The program also synchronized the EMG signal with the accelerometer signals from the shaker control unit. Finally all data were stored in a database. As in studies I and II the RMS value was normalized against the MVE and presented as % MVE.
<u>Study V.</u> EMG analyses were performed with the Mega Win 2.0 software (Mega Electronics Ltd, Kuopio, Finland) using a Hanning window of 1s prior to the FFT. The slope of the median frequency change (ΔHz/s) was calculated in the Mega Win program for the first 15, 30 and 45 s, respectively, using a linear regression analysis. The initial

median frequency (IMDF) (Hz) was calculated as the intercept of the slope of the whole contraction time (45 s).
The normalized slope (Δ%/s) was calculated as the slope/IMDF*100. Analysis was performed using the mean of left and right side muscles (Larevière et al, 2002) to minimize variation. During neck flexion the sternocleidomastoid was analyzed and during neck extension the upper and lower dorsal neck muscles.

2.3.3 Biomechanical calculations

In study II, the sagittal static load moment (M) about the bilateral axis of the cervicothoracic motion segment C_7-T_1 was calculated using static biomechanical analysis:

$$M = d_h \times F_h + d_{eq} \times F_{eq},$$

where
d_h = moment arm from bilateral motion axis of C_7-T_1 to gravitational force of head-and-neck.
F_h = force induced by head-and-neck
d_{eq} = moment arm from bilateral motion axis of C_7-T_1 to gravitational force of head-worn equipment
F_{eq} = force induced by head-worn equipment weight

The load moment was induced by the mass of head-and-neck and: helmet itself, helmet and NVG (hNVG), and helmet, NVG and counterweight (hCW), respectively, with neck in neutral position and in 20° flexed position.
When the head is held in a neutral neck position, the neck is normally flexed about 10° in relation to the vertical plane (Harms-Ringdahl et al, 1999). Consequently, the neck angle for the flexed position was about 30° in relation to the vertical plane.
The distance between the motion axis of C_7-T_1 (measured at a point midway between the C_7-processus spinosus skin marker and the frontal cervical groove above the head of the clavicle (Harms-Ringdahl, 1986)) and the center of mass (just anterior to the tragus of the external ear (Dempster, 1955)) was measured to mean (sd) 157 (14) mm for the 14 subjects when sitting in a neutral position. To simplify the calculations the neck was considered as rigid and all flexion performed between C_7-T_1.
The head-and-neck was calculated as 7.9 % of body weight (Dempster, 1955) which for the pilots became 6.58 (0.57) kg.
The percentage of maximum voluntary contraction (% MVC) was calculated as the ratio between the load moments induced by the mass of the head and neck plus the head-worn-equipment and maximum strength multiplied by 100 (in study II presented as % MUR, but calculated the same way).
The pilots' isometric neck extension strength in a seated position, measured using a DBC140, was 48.7 (8.9) Nm.

2.3.4 Seat-to-head transmissibility

Figure 6. Tri-axial seat accelerometer positioned in helicopter seat.

Acceleration data was collected using six channels: one triaxial seat accelerometer (Endevco 2560 (Endevco Corporation, CA, USA) (pilots 1-13 in studies III and for all pilots in study IV), and a Brüel & Kjaer 4322 (Brüel & Kjaer, Nærum, Denmark) accelerometer (pilots 14-17 in study III)) placed on the seat (Figure 6), and three Brüel & Kjaer 4398 accelerometers mounted perpendicularly to each other in three directions (fore-and aft = X, lateral = Y, and vertical = Z) on an aluminum cube. The cube was attached to the helmet using a magnet glued to the cube (Figure 3) and two iron plates glued to the helmet. One was horizontal on top of the helmet and the other was inclined 20° backwards to correct for neck flexion (in study III). The acceleration signal was registered in the system controlling the shaker. Vibration transmissibility through the body was calculated as the ratio of results from helmet-mounted accelerometers and vertical vibration acceleration measured at the helicopter seat (Magnusson and Pope, 1998).

2.3.5 Subjective ratings

During the 45 s isometric contraction in study V the subjects rated their subjective fatigue from the neck region at 15 s, 30 s and 45 s, using the Borg category ratio scale (CR-10) (Borg, 1982). This is a numerical 10-point scale with verbal anchors placed at selected positions along it in such a way that the scale is suggested to acquire the quality of a ratio scale.

2.4 Statistics

2.4.1 Repeated measures

For this thesis the results from study I were reanalyzed using a mixed-model ANOVA with muscle activity for each muscle as dependent variables and subjects as random factors. In study I, head-worn equipment (3 levels), neck and body inclination (3 levels) and neck rotation (3 levels) were used as fixed factors. In the model the interaction between head-worn equipment and neck and body position was also included as a fixed factor.

In study I the EMG data was subtracted with the lowest detected value during the test. The drawback with this manipulation was that data often consisted of a zero-value, making logarithmic transformation impossible. When re-analyzing the EMG-data, the evaluation of residuals implied that the data should be log-transformed. For this reason, the statistical analysis presented here was performed on the log-EMG data without subtracted baseline.

For the EMG analysis in study III a two-way, mixed-models ANOVA was conducted using the mean muscle activity level during the five-minute test runs for each muscle as dependent variables, subjects as random factor and neck position (2 levels) and head-worn equipment (3 levels) as fixed factors. A second mixed-models ANOVA was calculated using neck position and head-worn equipment as fixed factors and subjects as random factor. As dependent variables, vibration transmissibility peak levels and location in fore-and-aft, lateral and vertical directions were used. In the models the interaction between neck position and head-worn equipment was also added as a fixed factor.

For the EMG analysis in study IV a mixed models ANOVA was conducted using the mean muscle activity level over the 5 min test runs for the respective muscle as dependent variables, vibration magnitude (3 levels) as fixed factor and subjects as random factor. A second mixed-models ANOVA was calculated using vibration magnitude as a fixed factor (3 levels) and subjects as random factor. As dependent variables vibration transmissibility peak levels and location in fore-and-aft, lateral and vertical directions were used.

A p-value below 0.05 was considered significant. When a significance was found on the main level, a post-hoc test (Tukeys LSD) was performed to evaluate differences within each factor.

2.4.2 Reliability

Prior to the statistical analyses in study V, all results were checked with Bland-Altman plots to make sure that the variance was not related to the mean.
One-factor repeated measures ANOVA was performed to test for systematic differences in the EMG parameters intra- and inter- days. From the ANOVA table, suitable factors were selected for the statistical analyses to assess the reliability of the four parameters slope, normalized slope, IMDF and Borg scale, using three types of statistical calculation:

(i) The intra-class correlation coefficient (ICC1.1) was calculated using the formula:

$$(BMS - WMS) / (BMS + (k - 1) * WMS)$$

where BMS is the variance between subjects, WMS the variance within subjects and k the number of measurements. For the ICC values it is current practice to accept 0.8-1.0 as ‘excellent repeatability’, 0.6-0.8 as ‘good repeatability’, while ICC values below 0.6 are ‘poor repeatability’ (Bartko, 1966).

(ii) The standard error of measurement (S_w) was calculated as the within-subject standard deviation (*i.e.* the square root of within-subject variance). To present a measure of repeatability (Bland and Altman, 1996) we calculated:

$$\sqrt{2} * 1.96\ S_w, \text{ or } {\sim}2.77 * S_w$$

where the difference between two measurements for the same subject is expected to be less than 2.77 S_w for 95 % of pairs of observations (Bland and Altman, 1996)

(iii) The coefficient of variation (CV) was calculated using the formula:

$$S_w / \text{mean} * 100$$

which demonstrates the S_w relative to the mean and is considered to allow for comparisons between different values and methods, and is especially useful if the data is heteroscedastic (see e.g. Atkinson and Nevill, 1998).

To assess for differences in Borg ratings, intra- and inter- day, a Friedman analysis was performed.

2.4.3 Correlation

To compare the different evaluation methods, Spearman correlation coefficients (r_s) were calculated for this thesis.
Induced load and muscle activity were compared in study II, where each individual had his own correlation coefficient.
To compare vibration transmissibility and muscle activity in study III, the RMS values for 1/3 octave frequencies were calculated beginning at 2.5 Hz. This gave the following intervals: 2.50-3.15 | 3.15-3.97 | 3.97-5.00 | 5.00-6.30 | 6.30-7.94 | 7.94-10.00 | 10.00-12.60 | 12.60-15.87 | 15.87-20.00 | 20.00-25.20 | 25.20-30.00 Hz, each covering 13 or 14 seconds. In the correlation analysis only the first 2.5 minutes of the tests was analyzed. As with the induced load, this analysis was also performed on each individual pilot.
In study V the correlation between the subjective ratings and the slope of the median frequency was analyzed after 15, 30 and 45 s on the four test occasions. Here the correlation coefficient was based on group data.

3 Results

For the analysis of muscle activity and vibration transmissibility there was no significant interaction effect of seat positions and head-worn equipment, and thus only the main effects are presented below.

3.1 Neck and body position

The results concerning neck load caused by different neck and body positions are based on the findings from studies I, II and III. For a schematic presentation of some of the results see Figure 7.

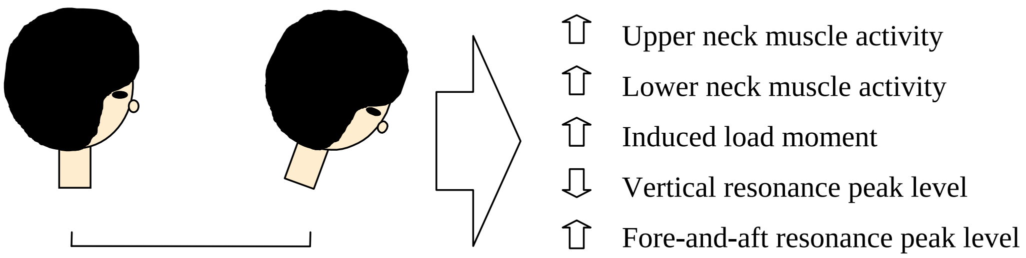

Figure 7. Schematic presentation of the change in measured variables when changing from neutral neck position to 20° neck flexion. Up-arrows indicate an increase in measured variable and down-arrows indicate a decrease.

3.1.1 Muscle activity as a function of neck and body position

Analysis of the logarithm of muscle activity data in study I revealed an increase in the upper and lower dorsal neck muscles as an effect of rotation ($p<0.001$), where the mean muscle activity during ipsilateral rotation was 2.7-4.1 % MVE higher than for neutral position. There was also a significant increase in upper and lower neck muscle activation as an effect of neck flexion and trunk inclination ($p<0.001$). The mean muscle activity during neck flexion was 1.7-3.0 (1.0-1.5 in study III) % MVE higher, and during trunk inclination 1.3-2.6 % MVE higher, than in neutral position.

3.1.2 Mechanical load as a function of neck position

The induced flexing load moment increased by about 4 Nm when the neck was flexed as compared to neutral position. This represents an increase in neck load by about 8 % MVC.

3.1.3 Seat-to-head transmissibility as a function of neck and body position

Neck position affected transmissibility substantially for the resonance peak level in the fore-and-aft direction: transmissibility with the head flexed was about 0.8 m/s^2 higher than when the head was in the neutral position (flexed: 1.9-2.2 m m/s^2; and neutral: 1.0-1.5 m/s^2, respectively), irrespective of head-worn equipment ($p<0.001$). In contrast, vertical transmissibility was higher in the neutral position (mean level 1.7 $m/s^{2)}$ than with the head flexed (1.5 m/s^2) ($p<0.001$). Transmissibility in the lateral direction was generally lower than in the other directions and the difference between neck positions was smaller (flexed: 0.8 m/s^2, neutral 0.7 m/s^2) (p=0.002).

3.2 Head-worn equipment

The results concerning neck load caused by different head-worn equipment are based on the findings from studies I, II and III. For a schematic presentation of the results concerning effects of NVG, see Figure 8.

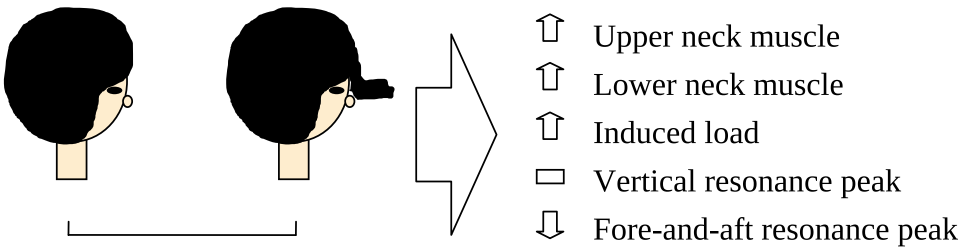

Figure 8. Schematic presentation of the change in measured variables when adding NVG to the helmet.

3.2.1 Muscle activity as a function of head-worn equipment

In studies I and III the muscle activity increased in the upper and lower dorsal neck muscles when wearing hNVG and hCW compared with wearing helmet only (p<0.05). The exception was the left and right lower neck using hNVG in study I (p=0.079 and 0.325, respectively).

The increase in mean muscle activity when using hNVG and hCW compared to helmet alone was 1.6 (in study III 0.7) and 1.7 (0.5) % MVE, respectively, in left upper neck and 0.6 (0.6) and 0.7 (0.5) % MVE, respectively, in right upper neck.
In the left lower neck, the increase was 0.8 (0.6) and 1.7 (0.5) % MVE and in right lower neck 0.4 (0.6) and 0.8 (0.5) % MVE. For the trapezius muscle and sternocleidomastoid there was no apparent effect of the different head-worn equipment.

3.2.2 Mechanical load as a function of head-worn equipment

The induced mechanical load calculated about the bilateral motion axis of C_7-T_1 increased by 1.3 Nm when NVG were used compared with helmet only in neutral position and by 1.6 Nm in flexed position, i.e. an increase in induced load by 2.6 and 3.4 % MVC, respectively. The counterweight reduced the induced load by 0.6 Nm in neutral position and by 0.3 Nm in flexed position, representing decreases in induced load by 1.2 and 0.7 % MVC respectively.

3.2.3 Seat-to-head transmissibility as a function of head-worn equipment

The fore-and-aft transmissibility response when using different head-worn equipment differed significantly in resonance peak concerning level (p<0.001) and location (p=0.016): the helmet peak levels were higher and located at lower frequencies. For the other peaks there were no significant differences between any of the head-worn equipment sets.

3.3 Whole-body vibration

The results concerning neck load caused by whole-body vibration are based on the findings from studies III and IV.

3.3.1 Vibration frequency

3.3.1.1 Muscle activity as a function of vibration frequency

The median activity in upper and lower neck muscles during the six tests in study III is presented in Figure 9 using a five-second moving average. Visual inspection of the muscle activity charts indicates that the activity increased by about 0.5-1 % MVE in the upper neck muscles at frequencies around 5 Hz for all variations of head-worn

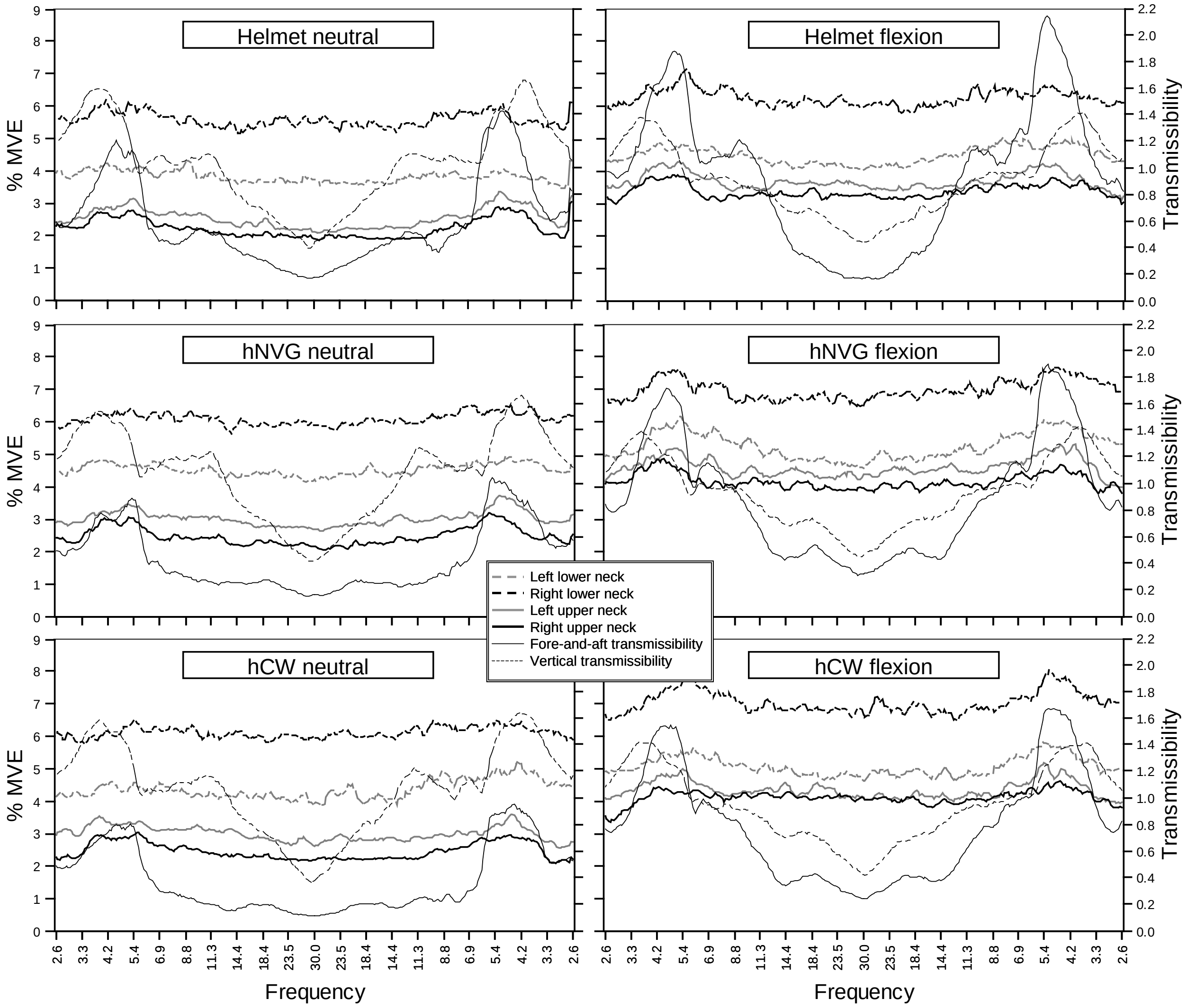

Figure 9. Median muscle activity and seat-to-head transmissibility (n=17) during a 5-min sine sweep with three different head-worn equipment sets (hNVG = helmet and NVG and hCW = helmet, NVG and counterweight) and two neck angles.

equipment and neck position. For the lower neck muscles, activity increased by about 1 % MVE at frequencies around 5 Hz when the head was flexed.

3.3.1.2 Seat-to-head transmissibility as a function of vibration frequency

The median transmissibility responses to the vibration in X and Z direction for all six test runs in study III are presented in Figure 9 using a five-second moving average. A screening of the individual data indicated four peaks in the X and Z directions. The first and most significant (resonance) peak was in the range 2.5-6 Hz, the second between 6 and 10 Hz, the third between 10 and 15 Hz and the fourth between 15 and 22 Hz. For the Y direction, only one peak was distinguished, ranging from 3-7 Hz.

3.3.2 Vibration magnitude

3.3.2.1 Muscle activity as function of acceleration magnitude

The median muscle activity in upper and lower neck muscles during the three tests in study IV is presented in Figure 10 using a five-second moving average. Visual inspections of the muscle activity charts indicate that the activity was basically unchanged at the 0.5 m/s^2 and 1 m/s^2 magnitude for all muscles analyzed. At the 2 m/s^2 magnitude the muscle activity increased at the lower frequencies, with a peak at about 4-5 Hz. The response to the vibration magnitude also differed between right and left muscles where the right trapezius and right lower neck muscles were more affected by the vibration than the left side. In the upper neck the response was the opposite, with higher levels in the left muscle. In the sternocleidomastoid there were no apparent side difference.

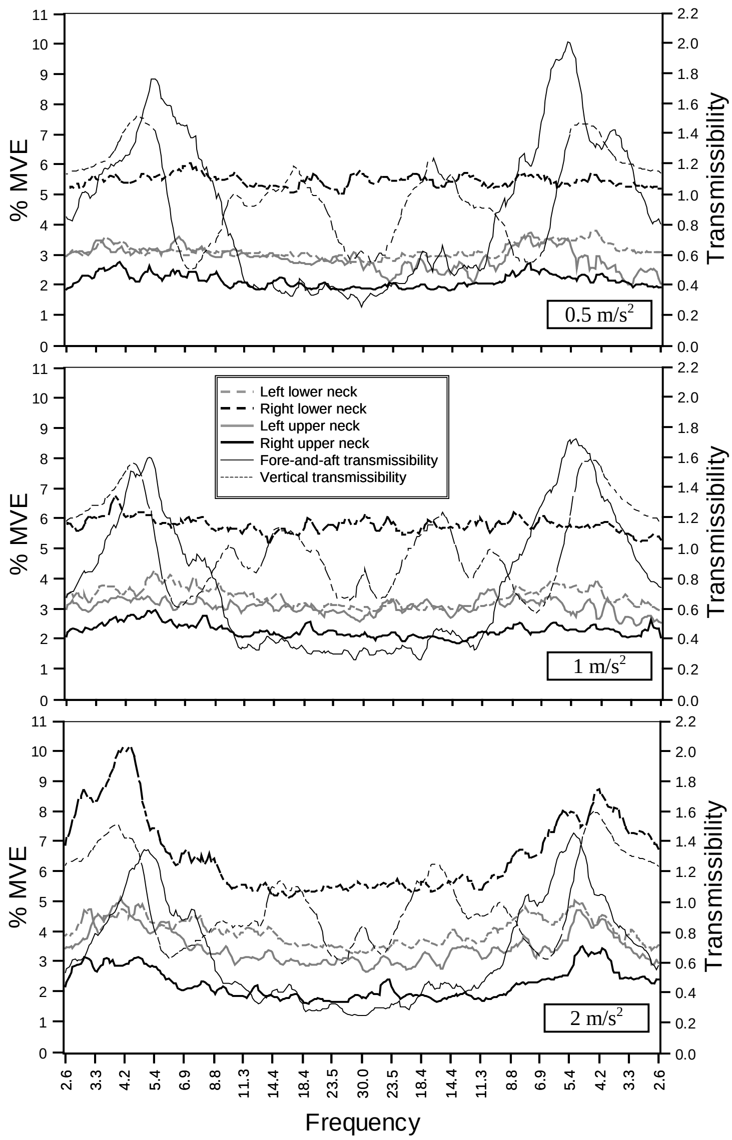

Figure 10. Median muscle activity and seat-to-head transmissibility (n=10) during a 5-min sine sweep at three different vibration magnitudes.

Statistical analysis of the mean muscle activities during the whole 5-min test runs revealed a significant effect of acceleration magnitude in both left (p=0.003) and right (p=0.001) lower neck and left (p=0.007) and right (p=0.011) trapezius muscles, where the 2 m/s^2 magnitude was higher than the other magnitudes. There was no significant difference between the two lower vibration magnitudes.

3.3.2.2 Seat-to-head transmissibility as a function of acceleration magnitude

The median transmissibility responses in X and Z directions to the vibration for all three test runs in study IV are presented in Figure 10 using a five-second moving average. Screening the individual data indicated one peak in X and Y directions ranging from 3-7.5 Hz and three peaks in Z direction. The first (resonance) peak was in the range 2.5-7 Hz, the second between 7 and 12 Hz, and the third between 13 and 20 Hz. Peak analysis revealed a distinctive first peak during every individual test run, in vertical and fore-and-aft direction, and for about half the occasions in lateral direction.

There was a significant difference in resonance peak level in fore-and-aft direction (p=0.04) where the highest vibration magnitude caused the lowest transmissibility level. For the lateral and vertical directions there were no difference between vibration magnitudes (p=0.928 and 0.856, respectively). The frequency of the principal resonance peak decreased both vertically (p=0.006) and fore-and-aft (tendency, not significant though, p=0.077) as the vibration magnitude increased.

3.4 Reliability

3.4.1 Slope of the median frequency

The results for the median frequency slope showed generally better reliability parameters the longer the contractions lasted. The highest ICC value combined with the best (i.e. lowest) S_w was obtained for upper neck, intra-day, on the 45 s contraction. Here, the ICC indicating 'excellent' repeatability. For lower neck, the intra-day, 45 s comparison also presented the best reliability, but with somewhat poorer results than in upper neck. ICC indicated 'good' repeatability. The ICC values for the lower neck were consistently lower than for the other electrode locations, even though in many cases the S_w was the lowest. The low ICC values thus indicated small between-subject variance.

The ICC values for the sternocleidomastoid indicated 'good' to 'excellent' repeatability for all measurement occasions, but the S_w was higher than for upper and lower neck, indicating large variance between, as well as within, subjects. For all measured locations the greatest improvements were from 15 s to 30 s. From 30 s to 45 s, there were just small improvements in the reliability parameters.

3.4.2 Normalized slope of the median frequency

Of all comparisons, the upper-neck, intra-day, 45 s contraction was the most reliable concerning high ICC/low S_w. For lower neck, this was also the best combination, with somewhat worse results than for upper neck. However, the lowest S_w for the lower neck was obtained during the inter-day, 45 s contraction. For this analysis, the ICC was only 0.09, because of a small between-subject variance. For the sternocleidomastoid there were generally high ICC values, and the S_w improved the longer the contraction was performed. As for the slope (Hz/s), the difference between 30 s and 45 s was small.

3.4.3 Initial median frequency

The results for initial median frequency (IMDF) showed generally low S_w values (CVs between 3 and 7 %) and ICC values indicated ‘good’ to ‘excellent’ intra-day repeatability. For the inter-day analysis the lower neck values showed high repeatability with high ICC/low S_w, whereas the other two electrode locations analyzed showed low ICC values, indicating poor repeatability.

3.4.4 Subjective ratings

In contrast to the variables considered above, the subjective rating of perceived fatigue showed the best reliability coefficients for the 30 s inter-day calculation. The only analyses that could be considered reliable for the subjective ratings were, for flexion, inter- and intra-day up to 30 seconds and, for extension, inter-day for 30 s and 45 s. Noteworthy are the high rating levels after 45 s, with a mean in some cases over 9 out of 10.

3.5 Comparison of evaluation methods

3.5.1 Correlation between EMG and induced load

The correlation between muscle activity (% MVE) and induced load (% MVC) in the upper and lower neck varied among the 14 pilots. In upper neck the median r_s was 0.87 with a range of 0.33 to 0.99. The correlation for the lower neck was generally somewhat lower with a median r_s of 0.69 and a range of 0.12 to 0.93. Notable is that at two of the lowest r_s for the upper neck (0.33 and 0.52) the correlation for the lower neck was high (0.83 and 0.80).

3.5.2 Correlation between EMG and transmissibility

The correlation between muscle activity and seat-to-head transmissibility varied substantially between subjects, muscles and transmissibility direction (Table II). The median correlation was highest for the fore-and-aft direction for right upper and lower neck with r_s of 0.49 and 0.50 respectively. The median correlation coefficient was positive for all muscles and directions but lower than for the two mentioned. However, every muscle and direction there were pilots who had negative correlations indicating large individual variation.

Table II. Spearman correlation coefficients for seat-to-head transmissibility versus muscle activity in upper neck, lower neck, the sternocleidomastoid and the trapezius (n=17).

	Fore-and-aft		**Vertical**		**Lateral**	
	Median	Range	Median	Range	Median	Range
Left upper neck	0.24	-0.31 - 0.89	0.24	-0.22 - 0.86	0.14	-0.55 - 0.62
Right upper neck	0.50	-0.08 - 0.91	0.37	-0.23 - 0.94	0.32	-0.49 - 0.85
Left lower neck	0.37	-0.48 - 0.94	0.37	-0.51 - 0.90	0.24	-0.53 - 0.82
Right lower neck	0.49	-0.13 - 0.95	0.37	-0.38 - 0.86	0.18	-0.33 - 0.75
Left sternocleidomastoid	0.31	-0.47 - 0.84	0.19	-0.54 - 0.90	0.17	-0.67 - 0.62
Right sternocleidomastoid	0.26	-0.40 - 0.89	0.07	-0.40 - 0.79	0.08	-0.48 - 0.95
Left trapezius	0.25	-0.84 - 0.85	0.32	-0.77 - 0.79	0.19	-0.81 - 0.88
Right trapezius	0.19	-0.43 - 0.85	0.25	-0.37 - 0.85	0.11	-0.43 - 0.65

3.5.3 Correlation between EMG and subjective ratings

The correlation between the slope of the median frequency and the subjective ratings was significant only for upper neck after 15 s in the second test, on day one and after 30 s on day three (Table II). The Spearman's r_s values were for the upper neck all negatively correlated (-0.18 to –0.84). For the lower neck and the sternocleidomastoid, the results were inconsistent, with both positive and negative r_s values (Table III).

Table III. Spearman correlation coefficients for subjective ratings versus slope (ΔHz/s) of median frequency for upper neck, lower neck, and sternocleidomastoid, after 15, 30 and 45 s for all test occasions.

Variable	**Day 1.1**	**Day 1.2**	**Day 2**	**Day 3**
Slope ΔHz/s				
15 s				
Upper neck - Extension	-0.23	-0.77	-0.50	-0.64
Lower neck - Extension	-0.58	-0.14	-0.10	-0.47
Sternocleidomastoid - Flexion	-0.48	-0.15	0.18	-0.38
30 s				
Upper neck - Extension	-0.25	-0.39	-0.50	-0.72
Lower neck - Extension	0.19	-0.25	-0.33	0.03
Sternocleidomastoid - Flexion	-0.02	0.10	0.03	0.19
45 s				
Upper neck - Extension	-0.20	-0.59	-0.38	-0.47
Lower neck - Extension	0.47	0.04	0.08	-0.02
Sternocleidomastoid - Flexion	0.10	-0.45	0.41	0.24

3.6 Other findings

3.6.1 Side difference

In studies III and IV the muscle activity in the right lower neck was constantly about 2 % MVE higher than left-side activity was (Figures 9 and 10). The absolute difference between left- and right-side trapezius was about 0.5 % MVE. In study IV there was also a difference in upper neck muscle activity, where the activity in the left muscles was higher than in the right side muscles.

3.6.2 Seat response to vibration

Another finding in studies III and IV was that the helicopter seat itself (with a seated person) had a resonance frequency in the same region as that of the head-and-neck (Figure 11). In the two studies two different chairs were used, one from a helicopter 9 (study III) and one from a helicopter 3 (study IV); but there were only minor differences between the two seats.

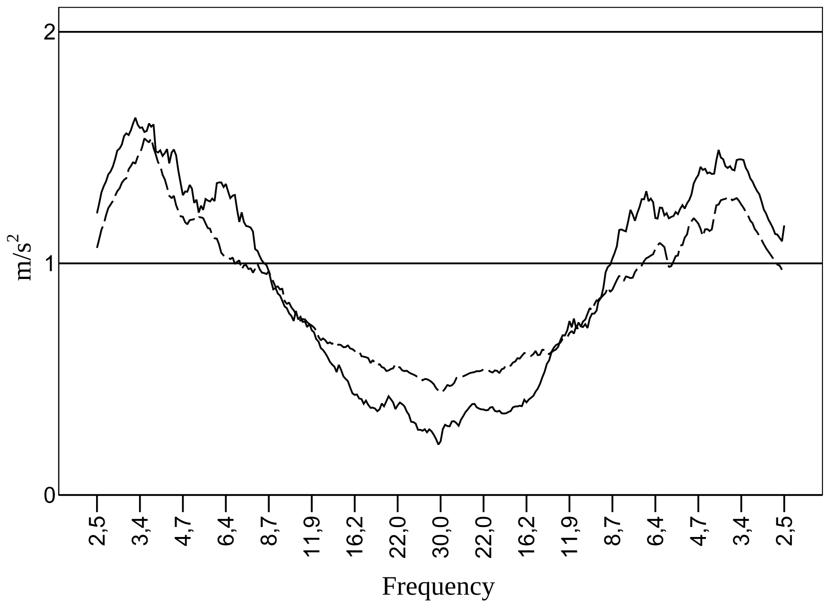

Figure 11. Seat acceleration at 1 m/s^2 for the two seats used in study III (broken line) and study IV (whole line)

4 Discussion

This present work sought to evaluate neck load among helicopter pilots as a consequence of different proposed risk factors. To evaluate the single risk factors, and also interactions thereof, all the studies were performed in a strict laboratory setting. The results can hopefully be used as a reference in future studies concerning neck load during flight. In physiotherapy it is essential to understand the possible causative factors that might contribute to load-elicited pain from neck structures. This is important in order to find the right intervention strategy for the specific group and to be able adjust the working environment where needed.

4.1 Discussion of results

4.1.1 Neck and body position

The results concerning neck and body position lead to a conclusion that, of the three risk factors evaluated, non-neutral neck position causes the greatest neck load measured as muscle activity, mechanical load and seat-to-head transmissibility of vibration. Another aspect of the sitting position is that the position in itself in the helicopter seat causes bilateral differences in muscle activity.
Sitting posture is also important, considering that cervical position depends on this (Black et al, 1996). A slouched sitting position, similar to the position assumed in the helicopter seat causes a higher cervical base inclination, which in itself increases neck load.

Throughout the projects reported in this thesis the pilots assumed a static position to minimize variation. During flight, however, there are needs for head movements, both vertical and horizontal. Head movements during flight have been evaluated in a few studies (Rostad et al, 2003a; Rostad et al, 2003b; Verona et al, 1986). During most of the time the pilot sits with his head in a fairly neutral position, no matter what head-worn equipment is used.

In study I, muscle activity during neck and body inclination, neck rotation and combinations thereof was evaluated. These positions are not generally assumed statically for long periods, but the results might give an indication of the load caused by non-neutral neck positions.

4.1.2 Whole-body vibration

In this work, whole-body vibration consisted of a sinusoidal signal, which is a simplification of the actual vibration in the helicopter. Since vibration magnitude and frequency in the helicopter depend on the number of rotor blades and their speed, the results may indicate what frequencies should be avoided.

It is evident that an increase in vibration magnitude affects the human body. The spine is well designed to counteract vertical gravitational forces. However, forces acting on the spine in other directions, or when the spine is in a non-vertical attitude, can cause shear forces, which must be counteracted by other structures than the spine. In study III we found that muscle activity increase at the resonance frequency (4-5 Hz) was more related to fore-and-aft head acceleration than to vertical transmissibility. The relative motion between the head and the body increases with higher vibration magnitudes even though seat-to-head transmissibility remains the same. For this reason, it is not surprising that the muscle activity increased with vibration magnitude.

The previously noted side difference due to seating position was also attenuated at the highest vibration magnitude. The right trapezius and lower neck and left upper neck showed markedly higher EMG activity at the resonance frequency than the other side did. Together with the results from study III, it seems that the vibration at a frequency around the principal peak resonance frequency increases muscle activity more if the muscle is already activated above 'base' level.

4.1.3 Head-worn equipment

The evaluation of head-worn equipment aimed to describe the effects of using night vision goggles in addition to the helmet. Of interest was also to see whether the use of a counterweight could reduce the load on the neck structures. Two different NVG equipment sets were evaluated. They differed somewhat concerning center of mass, but had similar masses. The muscle activity increase was somewhat lower in study III, which could be because the battery pack was mounted on the back of the helmet and thus better balanced. It is hard to compare the different results, though, since study I presents a mean of two seconds in nine test positions and study III presents mean muscle activity during the five-minute sweep.

When adding NVG to the helmet, the muscle activity increased, as did the induced load moment. In contrast, the seat-to-head transmissibility at the resonance frequency in fore-and-aft direction decreased. The additional use of a counterweight did not affect the results in any of the studies, even though the pilots reported a reduced load on the neck. These results could indicate the following; the counterweight weighs too little to give any great benefit, or; the objective evaluation methods used are too insensitive to detect the load-reducing capacity of the counterweight. Thus, any conclusion as to whether to use counterweights cannot be drawn from these results: the decision must be made by each pilot. From a biomechanical point of view it is recommended that the load moment of head-worn equipment should be below 8.28 Nm relative to the atlanto-occipital complex (Butler, 1992).

4.1.4 Neck muscle fatigue

The protocol evaluated in study V was constructed primarily for use in assessing neck muscle fatigue before and after flight. It can be discussed whether such an approach is possible, since it is likely that the mechanism causing muscle fatigue differs between the relatively low load during flight and the 75 % MVC contraction used in study V. A fatigued muscle would also have a lower initial median frequency, and an evaluation of the slope would thus give misleading results. However, Elfving et al (2003) reported

recovery of the median frequency of the back muscles within 5 minutes, and the time between landing the helicopter and taking the test is likely to be more than 5 minutes. A similar approach as in study V was used by Phillips and Petrofsky to evaluate effects of different head-worn equipment weights on neck muscle fatigue and endurance (Phillips and Petrofsky, 1983a; Phillips and Petrofsky, 1983b). They reported a greater decline in the median frequency of the cervical muscles during static contractions after exposure to a heavier combination of head-worn equipment than after exposure to a lighter one.
In addition to the issue discussed above, the protocol evaluated can also be used to evaluate neck muscle fatigability (Äng et al, 2005) and effects of different training regimes.

The overall best results were obtained after 45 s for all electrode locations. However, the target level of 75 % was quite strenuous to maintain during the whole contraction time, especially in flexion. Therefore it is suggested that, in future studies, the contraction time could be shortened to 30 s, since even though the results were somewhat worse than after 45 s, they still showed acceptable reliability. The gains of shortening the contraction time will probably outweigh the loss in reliability parameters, since the muscles will not be so fatigued. This was also evident when analyzing the Borg ratings. After 45 s the rating mean approached 10 (i.e. close to maximum fatigue). If this protocol is to be used to evaluate fatiguing effects of flying a helicopter, there cannot be a “ceiling” effect of the fatiguing contraction, i.e. there must be a possibility for the pilots to be more fatigued. This is probably the case after 30 s where the mean ratings were between 7.29 and 8.1.

For the upper neck there was some correlation between the slopes and the subjective ratings. Earlier studies on low-back muscle fatigue have in some cases reported high correlations between the slope and subjective ratings (Dedering et al, 1999; Dedering et al, 2000; Kankaanpää et al, 1997) and in some cases weak correlation between the slope and the subjective ratings, but high correlation between Borg-scores and different force levels (Dedering et al, 2002). The subjects in this study was to few to draw conclusions regarding to what extent the correlation between parameters can be used, but it can, even if vaguely, be seen as a validation of the EMG parameters as an index of muscle fatigue among the study population.

4.2 Methodological considerations

4.2.1 EMG

First of all one must remember that muscle activity is essential for a healthy muscle, and is generally of benefit for the human being. However, muscle activity also plays a significant role in the development of musculoskeletal disorders. This is easy to understand for tasks demanding high force, but for low force requirements it is not as straightforward (Sjogaard et al, 2000). It has been proposed that during low force contractions, single muscle fibers are selectively activated, and pain may occur. For that reason, the relatively low loads registered in this thesis are by no means harmful during short time contractions, but can, if sustained during longer periods, be a risk for development of neck pain.

Muscle activity during sitting was generally low, with some variation depending on neck position, head-worn equipment and vibration. During flight helicopter pilots sustain not only external mechanical load and aircraft control forces, but they are also sustained to psychological stress and often a need for precision handling of the sticks. During flight the possibility to adjust sitting position is very limited, and especially during maneuvering a stick must be held in each hand. The combined effect of static positions and mental stress can be a contributing factor for neck pain, especially since mental stress itself is known to increase muscle load (see e.g. Sjogaard et al, 2000).

The maximum voluntary contraction is limited by factors such as lack of motivation, pain, non-optimal joint angle and force direction. In this thesis the pilots were free of neck pain during the tests, and thus pain should not have been a problem. During the maximum neck extension and flexion contractions the applied force was monitored and thus the force level could be compared. This gave a hint as to whether the contraction was approximately maximal. Since the primary muscle groups of interest were the upper and lower dorsal neck extensors, the maximum contractions for these muscle groups was performed through neck extension in a neutral position, even though it fully possible to find neck angles that would generate a higher muscle activity.
There can also be ethical considerations when performing maximum contractions, especially for such vulnerable body parts as the neck. The participating pilots' neck strength was tested during their regular health check-up and they were thus familiar with the method.

With surface electrodes there is always a risk that EMG recordings include electrical signals from muscles other than the one being studied (Turker, 1993). For this reason, wire electrodes have been suggested as more accurate for selective recording of neck muscle activity (Mayoux-Benhamou et al, 1997). However, the present interest was not the selective muscle activity but, rather, activity from more widespread areas defined as the upper and lower neck extensors, respectively. Our electrode placements were the same as those defined in earlier studies (e.g. Schüldt and Harms-Ringdahl, 1988): splenius capitis and cervical/thoracic erector spinae, respectively. Probably, therefore, most of the EMG-signal recorded from our "upper neck" electrode came from the splenius capitis muscle and most of that for the "lower neck" from the cervical/thoracic erector spinae and the rhomboids.

Another problem with surface electrodes is that the skin may slide from its original position, leading to changed electrode placement in relation to the muscles of interest. This is generally not a problem when the head is in a neutral position, but is hard to control in flexed and rotated positions. This does not change the conclusions from comparing different types of head-worn equipment in each single position but care is needed when comparing rotated and flexed positions with neutral ones.

In studies III and IV the right-side lower-neck and trapezius muscles were more active than the left. This is probably because the non-symmetrical sitting position causes a relative right rotation of the neck, or because of the non- symmetrical arm and body positions. This is of great interest, especially for long flight missions. During flight there is also some requirement for forces applied to the sticks (Hewson et al, 2000b). This could further increase muscle activity levels and attenuate the side difference in muscle activity levels.

4.2.2 Biomechanical calculations

The biomechanical calculations in study II were performed using a static sagittal model. Since the cervical spine is far more complex, the results should be interpreted cautiously. In the model all the flexion was performed between the C_7-T_1 and the rest of the cervical spine was considered as rigid, which is a simplification of actual neck movements. However, measurements of neck load during flexion show that the major part of the head and neck load is induced at C_7-T_1 (Finsen, 1999). Even though this method is quite simple, the results are in line with those of other studies (Finsen, 1999; Harms-Ringdahl et al, 1986).

4.2.3 Seat-to-head transmissibility

In the two studies of seat-to-head transmissibility of vertical vibration, the accelerometers registering head movements were mounted on top of the helmet. Depending on helmet fit, unwanted motion will occur between head and helmet. In the two studies, two helmets were available, sizes medium and medium broad, and the pilots could use either. The helmets also have pads of different sizes, which can be adjusted to achieve best possible fit, thus minimizing head-helmet motion.
In a review of seat-to-head transmissibility, Paddan and Griffin (1998) found 46 studies concerning response to vertical vibration. Of these 14 evaluated transmissibility on top of a helmet or head-mounted sling. Most common was a bite bar, used in 21 of the studies. However, since neck muscle activity here depends on jaw muscle activation (So et al, 2004), such a method might influence the results when evaluating neck load using EMG.

As noted under Results, the seats themselves responded to the vibration in a similar way as the human body. If the only interest were to explore seat-to-head transmissibility, the best approach would probably have been to use the seat accelerometer as control accelerometer. However, our experiments were set to be semi-realistic helicopter mock-ups, and thus the differences in seat performance were part of the protocol.

4.2.4 Subjects

The targeted study population was Swedish military helicopter pilots. Due to the non-random recruitment of subjects, the volunteers are likely to have differed somewhat from the rest of the target population concerning neck pain and flying time. Given the nature of the present work and the relatively homogenous group helicopter pilots comprise, this has probably not affected the result in any significant way.

The results reported in this thesis might be of interest for other occupations with similar sedentary work environments. Other occupations that might be of interest are forestry machine drivers, forklift drivers, miners, etc where whole-body vibration, non-neutral neck positions and extensive head-worn equipment are frequently encountered.

While this thesis concerns only the effects of different external factors on neck load, it is unavoidable to discuss their relation to neck pain. In association with study III, the helicopter pilots reported in an informative questionnaire on what factors in relation to flying they experienced as potentially harmful. Nine factors were listed: the pilots ticked for the risk factor and body part they judged as a potential risk (Figure 12).

Vilka faktorer i samband med flygning anser Du vara en betydande risk för uppkomst/förvärrning av besvär?

	Nacke	Bröstrygg	Ländrygg	Armar	Ben
NVG	☐	☐	☐	☐	☐
Räddningsväst	☐	☐	☐	☐	☐
Sittställning	☐	☐	☐	☐	☐
Vibration	☐	☐	☐	☐	☐
Temperatur	☐	☐	☐	☐	☐
Buller	☐	☐	☐	☐	☐
Uppdrag (lång tid)	☐	☐	☐	☐	☐
Uppdrag (svårt att utföra)	☐	☐	☐	☐	☐
Flygstol	☐	☐	☐	☐	☐

Figure 12. Questionnaire concerning risk factors for pain development when flying a helicopter.

Of special interest for this thesis is what risk factors were ticked for the neck. Table IV lists the number and percentage of pilots (n=17) who ticked the listed factor as potentially harmful to the neck. The results from this short survey confirm the risk factors stated elsewhere (Bowden, 1987; Thomae et al, 1998) even though the safety jacket is rarely mentioned in the scientific literature.

Table IV. Number and percentage of pilots reporting the listed factors as a potential risk factor for neck injury (n=17).

Factor	Number	Percentage
NVG	12	71 %
Safety jacket	11	65 %
Sitting position	6	35 %
Vibration	9	53 %
Temperature	2	12 %
Noise	0	0 %
Mission (long)	13	76 %
Mission (difficult)	5	29 %
Pilot’s seat	4	24 %

The neck posture has been highlighted as a risk when wearing over-burdensome head-worn equipment. Different neck strengthening and stretching programs have been proposed to straighten the body and neck and to minimize “helicopter hunch” (Chrisman, 1999). However, the main problem remains in the sitting position itself, which is a great limitation to these training programs. To be able to fully allow helicopter pilots to adjust their heads as they prefer, the helicopter controls should be further developed and the space above the pilot’s head must not be limited.

4.3 Recommendations and future studies

In the long term, to minimize neck (and back) problems, the helicopter cockpit should be redesigned. Today pilots sits statically in an asymmetric position, with very little possibility for adjustment. The development of e.g. excavators has shown that complex vehicle maneuvering can be performed with the driver sitting in a comfortable shock-absorbing seat using one, or two joysticks.
In the short term a few things can be done to minimize the risk of neck pain. First, it seems reasonable that the pilot should have well prepared neck muscles in order to decrease the relative load. This can be achieved with well-planned training regimes. In addition the seating position in the helicopter should be individually adapted so as to minimize unnecessarily unfavorable sitting positions. Further recommendations based on the present results are:

- Non-neutral neck positions should be avoided, especially when using night vision goggles.
- Head space in the helicopter cabin should be analyzed to ensure that pilots can assume as comfortable a position as possible (perhaps taller pilots should avoid some helicopter models)
- Helicopter vibration should be analyzed with a view to avoiding frequencies around the resonance frequency of the human body.
- Helicopter seats should be evaluated. The seats used in the present work were from older helicopters, but nevertheless, we need to know the frequency responses of the new helicopter 14 and 15 seats.
- The present counterweight gave minor relief concerning muscle activity, induced load or vibration transmissibility, and thus no general use of counterweights can be recommended.

To obtain reference values for proposed risk factors all the studies reported in this thesis were performed in a strict laboratory environment. Using the knowledge gained, it would be of great interest to perform studies in a flying environment, with the following possible aims:

- perform a thorough examination of pilots’ sitting positions in the helicopter and to evaluate effects of suggested improvements,
- evaluate the effect of night vision goggles concerning neck muscle activity and fatigue during flight,
- compare different flying missions,
- compare novices and experienced pilots.

Further laboratory studies as follow-ups might include:

- comparison of muscle activity during sub-maximal ‘push’ as compared to ‘pull’
- production of a force-EMG curve for the evaluated group.

4.4 Conclusion

- All three proposed risk factors caused measurable changes in muscle activity, induced load and seat-to-head transmissibility. Of the three, neck and body position caused the highest response.
- At low loads, as apparent in the present thesis, substantial increases in induced load moment are entailed, with only a comparatively small increase in neck muscle activation levels.
- Vertical transmissibility decreased with the neck flexed, but fore-and-aft transmissibility increased, as did muscle activity. This indicates that transmissibility should be evaluated in more than one direction.
- A higher magnitude of vibration showed minor effects on the seat-to-head transmissibility level but increased neck load measured as muscle activity. This indicates that more than one outcome measure should be used to evaluate effects of whole body vibration.
- The results of the fatigue protocol tested in study V showed sufficient reliability to warrant the use of the intra- and the inter-day comparisons in further research protocols.
- Even though the reliability coefficients generally showed the best results after 45 s, we still conclude that a 30 s contraction time is preferable. The reliability results were only slightly poorer, and since the subjective ratings indicate that after 45 s the fatigue was very close to maximum, a shorter contraction time might be better.

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