Seated postural reactions to mechanical shocks
Laboratory studies with relevance for risk assessment and prevention of musculoskeletal disorders among drivers

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To Elin, Ludvig and Edvin!
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Abstract

Professional drivers of off-road vehicles, driving on irregular terrain such as in forestry, agriculture and mining, are exposed to whole-body vibration and mechanical shocks. These driver groups have reported severe musculoskeletal problems in the spine, but the association to seated postural reactions is not fully understood. One assumption is that unexpected shocks may create excessive load on spinal joints. The driver’s posture and exposure to mechanical shocks are required to be included in work risk assessments, but muscle activity and body kinematics are not included. The overall aim of this thesis was to describe and analyse seated postural reactions to mechanical shocks and to evaluate measuring of seated postures with relevance for risk assessment and the prevention of musculoskeletal disorders among drivers.

The thesis includes four studies, all laboratory-based using a repeated-measures design. Postural reactions were recorded from 23 (Paper I) and 20 (Paper II & III) young, healthy male participants who were seated on a movable platform. The platform delivered mechanical shocks with peak accelerations up to 14 m/s² in lateral directions during different conditions. Furthermore, twenty participants (Paper IV) were tested by four testers for analysis of test-retest reliability within and between testers measuring seated postures. Kinematics were here detected by means of a motion analysis system (MoLab™) and described for the spine as angular displacements or range of motion (ROM) using a three-segment model of neck, trunk and pelvis (Paper I–III) and as a more specific model (Paper IV). Surface electromyography (EMG) was recorded bilaterally on the following muscles; trapezius upper part, upper neck, erector spinae and external oblique (Paper I–III).

The general findings show that EMG amplitudes normalised to maximum voluntary contractions (MVC) did not exceed 2% in the trapezius, 8% in the upper neck and erector spinae and 18% in the external oblique. The EMG amplitudes and the angular displacements in the neck were significantly reduced from the first compared to the fifth mechanical shock. Adding a cognitive task significantly increased angular displacements. The largest ROM with approximately 20° in each segment was found during a double-sided mechanical shock (shock that changes direction). The reliability within one tester measuring seated postures was mostly considered good and superior to the reliability between several testers, but still insensitive to changes of less than 10°.

Exposure to single-sided or double-sided mechanical shocks with accelerations up to 14 m/s² seem not to cause postural reactions to such an
extent that overload of muscles or joint structures should be expected. There seems to be a quick adaptation that causes an improved readiness. The external obliques were most active when restoring equilibrium and seem important for stabilising the whole spinal column. Stability training, in order to improve neuromuscular control of the external obliques could, therefore, be a possible recommendation. The angular displacement in the neck increases if the subject solves a cognitive task of why such activities should be avoided when driving in difficult terrains. Since accurate descriptions of the spinal posture seems difficult even when advanced technical equipment is used, simpler models seem more appropriate. The results show that postural control is maintained even when exposed to considerable mechanical shocks. On the basis of these results, there is no need to change established risk assessment models.
Svensk sammanfattning

Yrkesförare som kör på ojämnt underlag, till exempel inom skogsbruk, jordbruk och gruvdrift, utsätts för mekaniska stötar och helkroppsvibrationer. Dessa förare rapporterar och uppvisar belastningsrelaterade besvär i ryggraden, särskilt i området kring nacken. De mekaniska stötarna, vibrationerna samt obekväma arbetsställningar anges som orsaker till de belastningsrelaterade besvären. Ett antagande är att oväntade mekaniska stötar kan skapa hög belastning på ryggradens strukturer. Kunskapen kring sittande kroppliga reaktioner till följd av mekaniska stötar är begränsad och därför inte heller medtagna vid en riskbedömning av yrkesförarens arbetsmiljö. Däremot har kraven höjts om att kroppshållningen ska beaktas och att exponeringen av mekaniska stötar ska ingå i riskbedömningen. Det övergripande syftet med denna avhandling var att beskriva och analysera sittande kroppliga reaktioner orsakade av mekaniska stötar samt utvärdera mätning av hållning i sittande ställning med relevans för riskbedömning och förebyggande av belastningsrelaterade besvär bland förare.

Denna avhandling består av 4 delstudier, alla utförda i laboratoriemiljö. Deltagarna, 23 personer i studie I och 20 personer i studie II & III, var alla unga friska män utan besvär i nacke och rygg. Hos dessa undersöktes kroppliga reaktioner till följd av hastigt påkomna mekaniska stötar levererade av en rörelseplattform på vilken de satt. De mekaniska stötarna, med olika karakter och accelerationer upp till 14 m/s², skedde endast i sidled. De kroppliga reaktionerna mättes genom rörelsesensorer vilka fästs på kroppen för att beskriva positioner och rörelseutslag av olika kroppsdelar genom en rörlig tre-segments modell (huvud, bål och bäcken). Även muskelaktivitet i nacke- och bålmuskler undersöktes med hjälp av elektromyografi som registrerade elektrisk aktivitet i muskulaturen. I studie IV ingick 20 deltagare samt fyra testledare för att testa tillsättligheten vid mätning av hållning med en mer specifikt modell. Hållningen mättes med 5 rörelsesensorer där varje sensor angav en vinkel för just det segment på vilken sensorn var placerad på.

Resultaten visade att muskelaktiviteten i trapezius (kappmuskeln), vilken vanligen förknippas med smärtan i nacke, var under 2% av dess maximala förmåga. Ingen av de andra uppmätta bål eller nackmuskulerna utnyttjade mer än 20% av sin maximala förmåga. Såväl muskelaktivitet som rörelsereaktion minskade till följd av upprepade mekaniska stötar. Vid tillägg av en uppmärksamhetskraavande uppgift ökade rörelseutslagen i nacke. Dubbelsidiga mekaniska stötar, det vill säga stötar som byter riktning, skapade större rörelseutslag i de olika kroppsegmenten jämfört enkelsidiga
mekaniska stötar. Tillförlitligheten när en person mätte sittande hållning var mestadels bra men förändringar mindre än 10° visade sig vara svåra att påvisa.

Slutsatsen är att en exponering för enstaka eller ett fåtal mekaniska stötar med accelerationer upp till 14 m/s² inte tycks skapa kroppsliga reaktioner i sådan utsträckning att överbelastning av muskler och ryggradens strukturer bör förväntas. Den yttre sneda bukmuskulaturen (external oblique) var den mest aktiva muskeln vid återställningen av balans efter en mekanisk stöt och således viktig för stabiliseringen av ryggraden. Stabilitetssträning av denna muskel kunde därför vara en möjlig rekommendation. Det sker en tillvänjning av de kroppsliga reaktionerna vid upprepning av mekaniska stötar vilket resulterar i en bättre beredskap. Även om rörelsereaktionerna varit små påverkas de av storleken och komplexiteten av de påförda stötarna, samt ökar om en uppmärksamhetskrävande uppgift skall lösas samtidigt. En rekommendation kan därför vara att undvika uppmärksamhetstörande moment vid körning i ojämn terräng. Eftersom en noggrann beskrivning av ryggradens hållning i sittande tycks vara svår att åstadkomma även när avancerad teknisk utrustning används, förefaller enklare modeller vara mer lämpligt att använda. Det finns utifrån dessa resultat ingen anledning att förändra etablerade modeller använda för riskbedömning av yrkesförares arbetsmiljö.
### Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>A[8]</td>
<td>Eight hour daily exposure action value</td>
</tr>
<tr>
<td>BMI</td>
<td>Body mass index</td>
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<tr>
<td>BOS</td>
<td>Base of support</td>
</tr>
<tr>
<td>CNS</td>
<td>Central nervous system</td>
</tr>
<tr>
<td>COM</td>
<td>Centre of mass</td>
</tr>
<tr>
<td>DSMS</td>
<td>Double-sided mechanical shock (of high peak acceleration)</td>
</tr>
<tr>
<td>EMG</td>
<td>Electromyography</td>
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<tr>
<td>EO</td>
<td>External oblique muscles</td>
</tr>
<tr>
<td>ES</td>
<td>Erector spinae muscles</td>
</tr>
<tr>
<td>ICC</td>
<td>Intraclass correlation coefficient</td>
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<tr>
<td>IMU</td>
<td>Inertial measurement unit</td>
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<td>LBP</td>
<td>Low back pain</td>
</tr>
<tr>
<td>L-SSMS</td>
<td>Single-sided mechanical shock (of low peak acceleration)</td>
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<td>MSDs</td>
<td>Musculoskeletal disorders</td>
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<tr>
<td>MU</td>
<td>Motor unit</td>
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<tr>
<td>MVC</td>
<td>Maximum voluntary contraction</td>
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<tr>
<td>Rep</td>
<td>Repetition</td>
</tr>
<tr>
<td>RMS</td>
<td>Root-mean-square</td>
</tr>
<tr>
<td>RMSE</td>
<td>Root-mean-square-error</td>
</tr>
<tr>
<td>SD</td>
<td>Standard deviation</td>
</tr>
<tr>
<td>Sed</td>
<td>Lumbar load expressed in terms of daily compressive dose</td>
</tr>
<tr>
<td>SDC</td>
<td>Smallest detectable change</td>
</tr>
<tr>
<td>SEM</td>
<td>Standard error of measurement</td>
</tr>
<tr>
<td>SSMS</td>
<td>Single-sided mechanical shock (of high peak acceleration)</td>
</tr>
<tr>
<td>UN</td>
<td>Upper neck i.e. Splenius capitis</td>
</tr>
<tr>
<td>UT</td>
<td>Upper trapezius</td>
</tr>
<tr>
<td>VDV</td>
<td>Vibration dose value</td>
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<tr>
<td>WBV</td>
<td>Whole-body vibration</td>
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Definitions

Base of support (BOS): The area of the body that is in contact with the support surface (1).

Agonist: Prime mover, muscles that contracts to cause an action. The left side muscles in a SSMS from the participant’s right side in this study (2).

Antagonist: A muscle that has an action opposite to that of the agonist. The right side muscles in a SSMS from the participant’s right side in this study (2).

Central set: A modified neuromotor state due to changes in initial context (3). Readiness to respond, based on training, expectation, anticipatory control and prior experience.

Centre of mass (COM): The point that is at the centre of the total body mass (1).

Equilibrium: The condition in which all the forces acting on the body are balanced such that the COM is controlled relative to the base of support (4). In this thesis in a neutral seated posture.

Mechanical shock: According to ISO 2041:2009 defined as a sudden change of force, position, velocity or acceleration that excites transient disturbances in a system (5). In this thesis refers to sudden accelerations, transmitted from the seat to a seated participant, causing disequilibrium.

Muscle synergies: The functional coupling of groups of muscles that are constrained to act as an unit (1)

Perturbation: A sudden change in conditions that displaces the body posture away from equilibrium (4). In this thesis refers to externally caused sudden changes of BOS. Synonymously used with mechanical shock.

Postural control: Control of the body’s position in space for stability and orientation (1).

Postural orientation: Ability to maintain an adequate relationship between different body segments and between the body and its surrounding (1).

Postural reaction: The purpose is to stabilize references values such as posture and equilibrium against internal and external disturbances (6). In this thesis the activity in response to a perturbation. In the literature also commonly referred to as postural response.
Postural stability (balance): The ability to control the centre of mass within the base of support (1). Provides the body carriage stability and conditions for normal functions in stationary position or in movement, such as sitting, standing, or walking.

Posture: The position or attitude of the body.

Whole-body vibration (WBV): According to directive 2002/44/EC defined as: “the mechanical vibration that, when transmitted to the whole-body, entails risks to the health and safety of workers, in particular, lower back morbidity and trauma to the spine” (7).
Original papers

The thesis is based on following publications, which will be referred to by their Roman numerals. Additionally, some unpublished data are included in the thesis.


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Preface

The reason why I became a PhD student are a result of two factors, firstly to adopt a new challenge and secondly to combine my two professions as a physiotherapist and mechanical engineer. I began training to become a physiotherapist in 2005, and had before that received a master of science in technology and worked as a design engineer in a medium-sized company with responsibility for quality, environmental and work environmental issues. The latter became a great interest where I wanted to learn more about the human body and work-related problems with the ambition of being able to help people and make a difference. After a while the interest had become so strong that I started my second profession. During the physiotherapy programme some new interests developed, but all the time there was a desire to reconnect with my previous profession and issues related to the work environment. After a few years working as a clinician, a project was announced related to the work environment with physiotherapy and technical skills desired. Since I like challenges, I applied and got it.

The focus of the project was to use novel inertial measurement units (IMUs) in order to study posture and postural reactions related to professional driving. It is important to be able to describe the body's posture and the postural reactions that occur during driving to detect possible causes of musculoskeletal disorders. Drivers often report spinal pain developed or made worse by driving for a long time with excessive exposure to whole-body vibrations (WBV), particularly to shocks and jolts, and also awkward seated postures. Although legislation exists for work risk assessments for the exposure of WBV and mechanical shocks, there is a lack of knowledge regarding the contribution from the active muscle reactions and body kinematics. At the start of my thesis a motion analysis system was developed (MT-R&D, University Hospital of Umeå) using IMUs. The idea was that the system should be able to measure both the mechanical shocks and body movements and thus obtain a better picture of the exposure situation. The IMUs have been further developed and improved during the thesis. I have had the possibility to influence and change the orientation of content in the thesis due to possibilities, limitations, new results and knowledge. With the great help of my supervisors, co-authors and colleagues I have had the opportunity to discuss and learn more about these issues and possible explanations. During the process I have developed a scientific approach and began to understand the complexity of the human response to work-related exposure. With this thesis I hope to add knowledge regarding seated postural reactions to sudden laboratory-induced mechanical shocks and contribute to the improvement of related work risk assessments.
Introduction

Work–related musculoskeletal disorders (MSDs) are major issues in public health (8-10). The World Health Organization has defined work-related musculoskeletal disorders as all musculoskeletal disorders that are induced or aggravated by work and the circumstances of its performance. The term musculoskeletal disorder is, according to Luttman et al. (11), defined as health problems of the locomotor apparatus, i.e. muscles, tendons, the skeleton, cartilage, the vascular system, ligaments and nerves. Pain, fatigue and discomfort are the most common symptoms associated with MSD (12). In 2012 the two most frequently reported health problems in the European workforce were back pain and muscular pains in shoulders, neck and/or upper limbs (9). Work-related MSDs are a cause of concern not only because of the health effects on individual workers, but also because of the economic impact on organizations and society. The actual cost may be difficult to assess and compare, but the cost of work-related neck and upper limb MSDs has within the EU been estimated to be between 0.5% and 2% of the gross national product (8). The annual prevalence of neck pain in the working population varies from 27% to 48% (13) with geographic differences, e.g. a higher prevalence in Scandinavian countries compared with other European countries (14). Low back pain (LBP) is a problem that most people experience at some point in their life (15). Worldwide, 37% of the LBP was deemed attributable to occupational risk factors (16).

Musculoskeletal disorders and seated at work

People are spending much of their time seated because of the change in modern working life. Too much sitting is adversely associated with health outcomes, including cardio-metabolic risks, which is why a limited sitting time is suggested (17). Sitting duration alone has been suggested not to be a causal factor in developing LBP (18-22), but there may be other explanatory factors, such as the posture adopted. Still, excessive periods sitting have commonly been reported as an aggravating factor for LBP (23, 24) and sitting time has also been found to be positively associated with LBP intensity (25). Epidemiological studies suggest that excessive sitting periods alone are an important risk factor for developing neck pain (13, 26-28), and a recently published study suggests an association between sitting time and neck and shoulder pain intensity (29). Also, work posture, e.g. excessive neck flexion, has been positively associated with neck pain (13, 26-28).

Spinal curvatures associated with seating posture have seldom been reported in previous epidemiological studies. One reason could be the lack of sensitive measurement techniques and thus the duration and frequency of different
postures has been more frequently investigated (e.g. sitting versus standing) (18, 20). The risk of musculoskeletal symptoms and MSDs may be reduced by encouraging specific seated postures (30). However, there is no consensus on an ideal seated posture (31-34). In general, favourable posture maintains spinal curvatures, retaining the joints in neutral positions (35, 36) to avoid excessive tissue strain that may cause MSDs (37). To adapt to a sitting position, the hip joint needs to flex, allowing the hip flexors to relax and the extensors, e.g. hamstring, to increase their tension. When the hip extensors stretch, they exert a pull on the pelvis, which causes it to rotate backwards and the lumbar curve to flatten. The rotation causes increased trunk moment with an increased intra-disc pressure. The pressure could also increase due to disc deformation (38). A more explicit rotation causing a flexed posture could also induce shearing forces to the spine that could cause LBP (39). Additionally, there are theories that suggests that a sedentary posture without continuous alternation could increase the disc load and also inhibit the intervertebral discs nutrition (19).

The seated posture depends on factors such as chair design, sitting habits, the task, seat inclination, backrest position, shape and inclination, and other supports. A change in the seated posture also influences the neck posture and alters the muscle activity (40, 41). The head-neck system is a complex biomechanical linkage with at least 20 pairs of muscles rendering a range of opportunities in stabilizing the head (42). The movements occur in the intervertebral joint and the two facet joints which makes the cervical spine stable and flexible (43). A sustained neck posture deviating from a neutral cervical spine, e.g. neck flexion or forward head posture, causing flexion in the lower cervical and extension in the higher cervical, has been suggested to cause static loading of the neck muscles that may cause neck pain (26, 40).

**Whole-body vibration and mechanical shocks**

The exposure to vibration in the European workforce varies between 5 and 25% depending on country and inclusion criteria (44) where exposure to whole-body vibration (WBV) occurs primarily in seated positions. Vibration arises when a body oscillates due to external or internal forces. Vibration will be transmitted to the human body through the part in contact with the vibrating surface, for WBV mainly from the seat and floor in a mobile machine. The reported number of off-road machines and tractors in each country within the EU varies between 100,000 and 150,000 for off-road machines and 300,000 and 1,400,000 for tractors (44). Off-road vehicles include quadbikes, excavators, snowmobiles and forest machines such as forwarders and harvesters. Driving such vehicles on irregular terrain will cause WBV that also includes mechanical shocks or vibration transients that
is non-stationary with high amplitudes that appear during a short period of time. Mechanical shock is thus a part of WBV, but with one major difference – that they only occur suddenly and without periodicity. Most research has been conducted on WBV without the contribution of mechanical shocks, or without distinguishing the contribution of mechanical shocks.

**Risk assessment**
The vibration is determined by the magnitude, the frequency and duration of the oscillation. The magnitude is in most cases expressed in terms of acceleration (m/s²), though displacement (m) and velocity (m/s) is also used. The acceleration can be expressed in terms of peak acceleration or root-mean-square (RMS) acceleration (22). The unit of frequency is Hertz (Hz) where 1 Hz is one cycle per second. WBV are evaluated in three orthogonal axes, fore and aft (x), lateral (y) and vertical (z). The human body response to WBV is not equal at all frequencies and directions. Frequencies below 20 Hz and accelerations in horizontal directions are considered more severe to the body response and are weighted and adjusted with a multiplied factor in such assessments (45). The orthogonal axes are combined by calculating a RMS acceleration that is used for exposure assessment. Exposure to WBV is assessed over an 8-hour period (A[8]) with fixed action and limits values (7).

The A[8] is at risk of being insensitive to mechanical shocks during long duration measurements, since their influence decays as the measurement time increases (46), but is still used in Swedish legislation AFS 2005:15 (47). When exposure includes mechanical shocks, different standards suggests different methods, like the crest factor (45), the Vibration Dose Value (VDV) used in the EU directive (7), or the internal lumbar load expressed in terms of a daily compressive dose, Sced (48). The latter aims to estimate the compressive load, i.e. the mechanical energy transmitted to the spine from the shock. The VDV is more sensitive to shocks and is a cumulative measurement of the vibration level received over an 8-hour period. One consideration is that all legislations are built on passive models without any contribution of body kinematics and muscle activity. One assumption in this thesis is that spinal injuries from mechanical shocks depend on two biomechanical phenomena. Firstly, an upright seated posture may become unstable if surrounding muscles are too passive. A sudden mechanical shock during such conditions may cause major local deformation in the spine before any active muscle reaction happens. This may cause larger deformation in the intrinsic structures than what is safe (49). Secondly, sufficient musculature activity together with passive and active stiffness of the spine may resist sudden mechanical shocks, but as a result may also cause high spinal loads that would be undesirable in the long term. Such loads may elicit loading above the tolerance limits of e.g. annulus fibres, endplate and ligaments (49, 50).
**WBV and mechanical shocks associations with work-related MSDs**

There is a consensus that exposure to WBV causes LBP and sciatica (51-54). The pooled odds ratio from a recent meta-analysis showed that WBV exposure doubled the risk of both outcomes (53). Also, adverse posture and prolonged sitting correlate with the exposure, but it is unlikely to explain the overall risk pattern (53). Older studies support a relationship between WBV and disc degeneration (52, 55), but a more recently published review found no causality between WBV and abnormal spinal imaging findings (56). The relationship between WBV and pathomechanism to LBP is, however, not clear (57, 58). One hypothesis is that the exposure to WBV causes mechanical overloads in the spine that is a potent disc herniation mechanism (59). Several cross-sectional studies also suggest that there is an increased risk of neck pain among workers who are exposed to WBV (60-65). For example, one meta-analysis supports the theory that the WBV exposure from vehicles is likely to increase cervical pathology among drivers compared to controls (odds ratio 2.33) (66).

It has been suggested that repeated exposure to sudden shocks would be more hazardous to the spine compared to continuous WBV exposure (54, 67-70) even though few studies have analysed and reported potential consequences. One meta-analysis by Water et al. (54) concluded that rough vehicle rides are prevalent and that repeated exposure to mechanical shock may increase the risk of LBP. The association between mechanical shocks and neck pain is less studied even though some reports exist (64, 69-71). Mechanical shocks might occur in all directions, but for drivers of certain vehicles such as forest machines and quadbikes, there can be a substantial exposure to WBV including mechanical shocks in lateral directions, e.g. Rehn et al. (71) and Milosavljevic et al. (72). Seidel et al. (73) concludes that more research into exposure in the horizontal directions are needed and suggests that posture and muscle activity should be included.

**Postural control**

Postural control is a complex skill and includes control of the body's position in space for the purpose of stability and orientation during any posture or activity (1). Postural orientation involves the ability to maintain the different body segments in relation to internal references, gravity, the environment and task (1). It is based on the central integration of vestibular, visual, proprioceptive and tactile information using feedforward and feedback. Postural stability involves the coordination of movement strategies to stabilize the centre of mass (COM) within the base of support (BOS) when the
equilibrium is disturbed for example by externally triggered perturbations (1). Postural control of the spine requires a complex interaction of musculoskeletal and neural systems in relation to the task being performed and the environment in which the task is being performed (1).

Local and global muscles are required for spinal stability. The local muscles, located deep within the body with attachments to the spine are responsible for control of the curvature and giving stiffness to maintain mechanical stability of the spine (74). The global muscles are large torque-producing muscles responsible for alignment, range of motion and balancing the outer load (74). Neither the local or global muscles in isolation can control stability. Panjabi conceptualized the spinal stabilization in three subsystems, the passive and active musculoskeletal subsystem and the control subsystem (neural) (75). A dysfunction in any part of the subsystems may cause spinal instability, a global increase in the movements associated with the occurrence of spinal and/or nerve root pain (76). There is an interaction between active (muscles), passive (spinal structures) and neural control subsystems (75) that have to work together for solving a postural task such as stabilized sitting. The spinal stability can be exposed to inadequate neural command, high repetitive low forces or low repetitive large forces (mechanical shocks) that may cause strain and sprain of spinal structures giving pain sensations (77). Prolonged loading occurs when seated and may in the short term affect functional responses in all subsystems, i.e. excitation/inhibition of sensory and reflexive autonomic reactions, recoverable fatigue of muscles, and creep or stress relaxation of discs and ligaments. Prolonged loading may in the long term lead to degenerative processes of the disc that could alter mechanical responses (49).

The active and passive musculoskeletal subsystem
The passive musculoskeletal subsystem includes the spinal column (vertebrae, intervertebral disc, spinal ligaments, facets articulations, joints capsules) and also the passive mechanical properties of the muscles. The active part consists of the muscles and tendons surrounding the spinal column. The passive subsystem does not provide any considerable stability in a neutral position, but provides reactive forces in the end of range of motion that resist further spinal motion. However, the passive subsystem acts as a transducer for measuring vertebral positions and a part of the neural control. The muscles and tendons in the active subsystem generates the forces required for spinal stability (75).

The activity in the muscles can be measured by electromyography (EMG). The muscles are composed of muscle cells, usually referred to as muscle fibres. The muscle fibres that are innervated by the same nerve (axon) are called motor units (MU) i.e. the smallest functional unit of a muscle (78). EMG provides an
indirect method to visualize activity in postural muscles and analyse the electrical signal derived from the MUs in the muscle when it is activated. Surface EMG is a non-invasive method where electrodes are attached on the skin. EMG gives information about the muscle and is composed of a network of active MUs. The EMG signal can be described as the amplitude of the signal by the root-mean-square (EMG RMS) or further normalized to a maximum voluntary contraction (% MVC). Also, the onset latency time i.e. the time from a stimulus to the first muscle reaction, can be estimated from the EMG signal.

**The control subsystem**
The central nervous system (CNS) controls postural stability through two types of strategies, open or closed loop. In open loop strategies, only feedforward input is used by CNS and occurs in, for instance, in pre-planned actions. In closed loop strategies, the reaction is generated by feedback, sensory input, from unpredicted disturbance or deviations from pre-planned movements (79). To detect the environment, the movement and the position, a multiple sensory system is needed. This includes skin mechanoreceptors, muscle receptors (muscle spindles), joint and ligament receptors plus visual and vestibular information. The system includes a broad spectra of activity from simple monosynaptic reflex to complex fine motor task (79). Between those extremes are the automatic reactions found in whole body perturbations. Automatic reactions are faster than voluntary reactions, but involve a more complex and widespread response i.e. muscles that are constrained to act as a unit (muscle synergies). Evidence suggests that the cortex is involved in altering these automatic reactions with changes in cognitive state, prior experience, initial conditions, and anticipation, all representing changes in the “central set” (80). The central set is defined as a modified neuromotor state due to changes in initial context (3). It is suggested that the cerebellum is responsible for adapting automatic reactions based on prior experience while the basal ganglia are responsible for pre-selecting and adjusting the automatic reaction based on context. Anticipatory postural adjustments serve to maintain postural stability by compensating for destabilizing forces associated with e.g. moving a limb, before they actually occur. During and after a perturbation, continuous feedback is used to adjust the reactions that may cause the central set to adapt. Automatic reactions are initiated, e.g. when the support surface in standing is rapidly moved in order to maintain the equilibrium of the body. The automatic reaction times when standing are suggested to be within 70–180 ms, longer than the stretch reflex latencies 40–50 ms and shorter than voluntary reaction times 180–250 ms (4). The reaction times when seated are not as well described in the literature.

All movements and postures involve a complex interaction of orientation and stability (6). The trunk includes half the body mass which means that trunk
movement is important for the control of postural equilibrium. The neuromuscular control of the head-neck system should ensure that the head is held steady for stabilizing the gaze (81). Since the head accommodates the visual and the vestibular systems, a postural perturbation that causes the head to move also causes these receptors to adapt and adjust to the new conditions. This indicates that postural control of the head should be a relatively difficult task and therefore important to study, where incorrect control may cause loads that results in musculoskeletal problems.

**Seated postural reactions**

Postural reactions due to external perturbation (mechanical shock) i.e. in which the Base of support (BOS) is suddenly moved, is rarely studied in a seated position compared to standing positions. The research carried out has mainly used translational perturbations in anterior-posterior directions (82-95), in some cases with other perturbation directions in additions to those (90-95). There are a couple of studies that have focused solely on lateral directions (96-98) and only one study that has used more complex perturbations (99). The initial detectable reaction due to perturbations occurs, due to inertia, closest to the contact point (87, 94, 100). Therefore, the movements start in the pelvis segment followed by the trunk and after that the head. It has been suggested that the head and trunk reactions initially rely on system mechanics and signals from segmental proprioceptors (101).

The muscle reaction from a perturbation has been shown to be direction-dependent in the trunk (91, 102, 103) and neck (94, 95). The muscle reaction to a lateral perturbation has been suggested to have a reciprocal activation pattern in the neck, starting in the contralateral muscle that first stretches (94, 96). The EMG amplitude in the neck region has been reported to be high, especially in the contralateral splenius capitis and is, therefore, most likely to be injured (96). Furthermore, the initial posture has been suggested to influence the nature of the postural reactions, e.g. a head rotation reduces the EMG amplitude in the upper neck (97).

Vibert et al. (94) have suggested two main categories of reaction strategies to lateral perturbations in seated positions; “stiff” and “sloppy”. The stiff strategy includes muscle co-contractions and the sloppy strategy a reciprocal, more relaxed muscle activity. The two specific strategies or a continuum in between those were found in the subjects tested. The individual chosen strategy was not found to adapt over time (94). A strict stiff strategy might cause increased load on the spine, muscle fatigue or myalgia in the long run. The relaxed strategy, depending on passive structures at the end range of ROM, might increase the risk for injury in tissues and cause pain sensations due to the
compression/stretching of neural elements or abnormal deformation in components in the passive subsystem (75). Contrary to Vibert et al. (94), who found no or little adaptation in the reactions to sideway perturbation, seated postural reactions to forward perturbation have been found to adapt after the first shock with decreased EMG amplitude (82, 88). The postural reactions could be affected by a cognitive task. Dual task studies of various kinds on standing, including balance recovery, have demonstrated impaired performance in one or both tasks and that cognitive processing, i.e. central set, is involved in controlling postural stability (104-106). The effect of a dual task on the upper body during perturbation in a seated position has not been studied.

Rationale for the thesis

WBV that includes mechanical shocks, for instance caused when driving on irregular terrain, is suggested to be more hazardous to the spine compared to WBV of a more stationary character, though few studies have analysed its cause and consequences. Drivers of off-road vehicles, exposed to considerable numbers of shocks, have reported severe musculoskeletal problems in the spine, especially from the neck region. However, the association between seated postural reactions caused by mechanical shock exposure and reported neck and back problems is not fully understood.

The mechanical shock exposure and the driver’s posture are suggested in international standards to be included in work risk assessments, but there is no clear consensus on how. One consideration is that all of the suggested work risk assessments are based on passive models without a contribution from any muscle activity or body kinematics. Whether the muscular activity and bodily movements arising from mechanical shocks in seated positions have the potential to increase the health risk is still unanswered.
Aims of the thesis

The overall aim of this thesis was to describe and analyse seated postural reactions when the individuals are exposed to mechanical shocks, and to evaluate technical measurements of seated spinal postures. This was done in order to acquire more information concerning the performing of work risk assessments when exposed to mechanical shocks and to suggest suitable prevention strategies for musculoskeletal disorders among drivers.

Specific aims

- Describe seated postural reactions in the neck and trunk in healthy men by using electromyography and inertial measurement units when exposed to mechanical shocks in lateral directions with a focus on the effects of various translational accelerations combined with or without a cognitive task. (Paper I)

- repeated mechanical shocks in lateral directions with a focus on the effects of different head postures and adaptation. (Paper II)

- single-sided and double-sided mechanical shocks in lateral directions with a focus on different perturbation characteristics. (Paper III)

- Evaluate the inter- and intra-tester reliability among testers using a system with inertial measurement units in order to measure spinal postures in healthy participants when seated. (Paper IV)
Methods

Participants

Young participants were targeted in order to reduce the risk of degeneration and rigidity of the spine (107) which could cause reduced mobility affecting the postural reactions (Table 1). The participants were primarily recruited among staff and students at Umeå University, Sweden. Exclusion criteria were any neurological conditions or reduced ability to perform daily routines during the last 12 months because of back or neck problems. Only males were included in paper I–III since the majority of professional drivers are men, but none of those included had worked as professional drivers. Four testers, one physiotherapist (TS), two physiotherapy students, and one biomedical engineer, were involved in the inter-tester reliability test (Paper IV). The intra-tester reliability test was performed by the same physiotherapist (TS) as the inter-reliability test. Paper IV was the first study in a chronological order, from which the results were used to secure the method of measuring seated participants before the main studies I–III were conducted. Paper IV included participants of both sex, but participants below a height of 160 cm were excluded as a consequence of the size of the measuring system units. Written informed consent was obtained from each participant. The studies were approved by the Regional Ethical Review Board in Umeå (No 2012-24-31M & Dnr 2014-228-32M).

Table 1. Characteristics of the participants. The participants in each data collection are unique. Paper IV included two groups of participants, one for the inter-test reliability and one for intra-test reliability, including different participants and a mixture of males and females (M/F).

<table>
<thead>
<tr>
<th>Paper</th>
<th>n</th>
<th>Sex</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>BMI (kg/m²)</th>
<th>Age (years)</th>
</tr>
</thead>
<tbody>
<tr>
<td>I</td>
<td>23</td>
<td>23M</td>
<td>181.0 (7.0)</td>
<td>79.3 (11.0)</td>
<td>24.1 (3.2)</td>
<td>24.2 (4.9)</td>
</tr>
<tr>
<td>II &amp; III</td>
<td>20</td>
<td>20M</td>
<td>180.6 (7.2)</td>
<td>79.3 (10.9)</td>
<td>24.5 (4.2)</td>
<td>27.5 (6.4)</td>
</tr>
<tr>
<td>IV Inter</td>
<td>10</td>
<td>8M/2F</td>
<td>178.9 (7.1)</td>
<td>74.5 (7.6)</td>
<td>23.3 (2.0)</td>
<td>27.6 (6.1)</td>
</tr>
<tr>
<td>IV Intra</td>
<td>10</td>
<td>6M/4F</td>
<td>175.7 (8.8)</td>
<td>75.9 (16.5)</td>
<td>24.3 (3.4)</td>
<td>30.5 (4.6)</td>
</tr>
</tbody>
</table>

Experimental procedure

A repeated-measures laboratory design was used for all studies. Laboratory studies were chosen so as to ensure that the measuring technique would work and so as to limit the number of influencing factors in the analyses. The procedures for paper I–III are presented thoroughly in Figure 1. A single-sided mechanical shock in a lateral direction of low peak acceleration (L-SSMS) or high peak acceleration (SSMS) was used in paper I. This was combined with or without a cognitive task (i.e. counting backwards in steps of
3 from a random number given by the test leader). The participants were allowed to practise once before measuring in order to reduce anxiety and lower the risk of quick adaption if the first reaction were to be divergent from the following reactions. The SSMS was further applied repeatedly with the participant seated with either neutral head postures (Paper II & III) or a laterally flexed head posture (Paper II). No practise was allowed before these measurements since adaption of the reactions was one of the aims and the first reaction of importance to measure. A double-sided mechanical shock (DSMS), e.g. a lateral mechanical shocks of high peak acceleration that changed direction with various time delays, fast, medium and slow, was used in paper III. All SSMS were delivered from the participant’s right side. The DSMS were delivered from the left side whereupon the direction change was determined by the second acceleration delivered from the participant’s right side (Table 2 & Figure 2).

**Figure 1.** Repeated single-sided mechanical shock in a lateral direction of low peak acceleration (L-SSMS) or high peak acceleration (SSMS) combined with or without a cognitive task. Repeated SSMS or double-sided mechanical shock (DSMS) changing direction with different time delays (fast, medium and slow). Participant seated with neutral (N), left (L) or right (R) laterally flexed head posture. The total number of repetitions (Rep) and the specific repetition (x-x) used for each analysis are defined. The mechanical shock and task (Paper I), the head posture (Paper II) and mechanical shock (Paper III) was partially randomised.
Table 2. Mean measured values of single-sided mechanical shocks of low peak acceleration (L-SSMS) and high peak acceleration (SSMS) and double-sided mechanical shocks (DSMS) changing direction with different time delays (fast, medium and slow). The accelerations (Acc 1 & Acc 2) were delivered from the participant’s right side (-) or left side (+).

<table>
<thead>
<tr>
<th>Shock (Paper)</th>
<th>Acc 1 (m/s²)</th>
<th>Acc 2 (m/s²)</th>
<th>Time to stop (s)</th>
<th>Stroke distance (mm)</th>
<th>Time to Acc 2 (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>L-SSMS (I)</td>
<td>-5.1</td>
<td>-1.2</td>
<td>240*</td>
<td>-</td>
<td></td>
</tr>
<tr>
<td>SSMS (I-III)</td>
<td>-13.3†</td>
<td>-0.8</td>
<td>240*</td>
<td>-</td>
<td></td>
</tr>
<tr>
<td>DSMS Fast (III)</td>
<td>14.3</td>
<td>-14.3</td>
<td>0.8*</td>
<td>90+180+90*</td>
<td>204</td>
</tr>
<tr>
<td>DSMS Medium (III)</td>
<td>14.0</td>
<td>-8.0</td>
<td>1.2*</td>
<td>110+220+110*</td>
<td>350</td>
</tr>
<tr>
<td>DSMS Slow (III)</td>
<td>15.6</td>
<td>-9.2</td>
<td>1.4*</td>
<td>140+280+140*</td>
<td>458</td>
</tr>
</tbody>
</table>

*Estimated values: †-13.2m/s² in paper I, - not relevant

Figure 2. Displacements of the movable platform by a single-sided mechanical shock in a lateral direction of low peak acceleration (L-SSMS), high peak acceleration (SSMS) or double-sided mechanical shock (DSMS) changing direction with different time delays (fast and slow). The time window epoch 1 (E1), used for EMG analysis, is displayed for the SSMS and the two DSMS.

All mechanical shocks were delivered by a 6 degrees of freedom movable platform, with an experimental flat chair, controlled by electrohydraulic actuators (Micro Motion System, Bosch Rexroth, Netherlands) (Figure 3). The participants were seated centred on the chair facing forward, in a good but relaxed sitting posture. Feet were placed together on a height adjusted foot rest so that the thighs were horizontal and foot contact could still be maintained. A cushion was placed between the knees and a belt buckled around the thighs. The hands were placed on the thighs with palms upwards so that all motor response as far as possible emanated from the trunk and neck.
Methods

**Figure 3.** The movable platform, with the experimental flat chair from the front and back with a participant seated equipped with inertial measurement units (IMUs) and electromyography (EMG).

In the fourth study, a written study protocol was followed four times by one and the same tester or once by four different testers. The tester mounted the inertial measurement units (IMUs), calibrated them, and gave instructions to the participant. The participants performed for each occasion, as similarly as possible, slow movement sequences including retained specific postures, while seated (Figure 4). The seated postures was inspired by Claus et al. (34). All postures were obtained 5 times per occasion and used for analyses. Practicing was allowed prior to the measurement being started. The IMUs was demounted between each occasion. The results were used to secure and adjust the method before the main studies I–III were conducted.

**Figure 4.** A) Lumbar kyphosis seated; B) Neutral posture seated; self-selected posture; C) Lumbar lordosis seated. Used in study IV.
Methods

Movement analysis

All kinematics were detected by an analysis system (MoLab™, AnyMo AB, Sweden) using 4–5 IMUs. Each IMU (MPU-9150 InvenSense, USA (Paper I–III) or Adis 16364 Analog Devices, USA (Paper IV)) included tri-axial accelerometers and gyros that detect accelerations and angle velocities used to calculate the relative and absolute angles of the body segments. Different sensors were used in the IMUs due to development of the system, but the sensors included had accelerometers and gyroscopes with at least 14-bit resolution and range of ±300 deg/s and ±10 g (≈100 m/s²). The angular resolution of the sensors was estimated to be approximately 0.06° at slow motions (108). Software calculated the real-time orientation and motion of the IMUs (108, 109). The system was sampled with ≥128 Hz and when needed synchronised with surface electromyography (EMG). In paper I–III the IMUs were placed on the seat, the back of the head and in the spinal processes at level Th2 and S2 using either adhesive tape or Velcro strap. Similar systems have been used and validated with good accuracy and inspired to the used setting (110, 111). The IMUs in paper IV were all placed on the body using the same positions and in addition two IMUs were placed on the spinal processes at level L3, and Th12 to differentiate movements in the lumbar spine (112). The IMUs in paper IV were attached on elastic Velcro straps and on a customized vest, which was chosen to be most feasible in field measurements and used previously in a study with a similar system (113). Movements within the spine were described as relative angles between two adjacent IMUs (Paper I–III). The peak angles, e.g. the maximum deviation from the neutral angle, or the range of motion (ROM), e.g. the deviation between two peaks of angle displacements where analysed in the frontal plane i.e. lateral flexion. There were three segments with relative joints; Neck (Head to Th2), Trunk (Th2 to S2) and Pelvis (S2 to seat). In paper IV, the position within the spine was described as absolute angles between the IMUs and its surrounding and analysed for each IMU in the sagittal plane i.e. flexion/extension where the major movements occurred.

Electromyography

Surface EMG was recorded with a sampling frequency of ≥1500 Hz, using the TeleMyo Direct Transmission System and model 542 DTS EMG Sensor (Noraxon USA Inc., US). The skin area of interest was shaved and cleaned with 70% alcohol solution before attaching bipolar circular Ag/AgCl electrodes (Ambu De) with an inter-electrode distance of 2 cm. Electrodes were placed bilaterally on the upper posterior neck muscles (UN), upper trapezius (UT), erector spinae lumbar level (ES) and external oblique (EO), described in detail in Table 3, using several references (114-116). All the muscles were selected on the basis that they could contribute to lateral flexion of the spine. Splenius
capitis, in this thesis referred to as UN, has been reported to be important in mechanical shocks in lateral directions (117) and with reported high amplitudes (96, 98). The trapezius has also shown relatively high amplitudes in mechanical shocks in lateral directions (96, 98) and has commonly been associated with pain in neck and shoulder (118, 119). ES and EO have previously been found to affect spinal stability with muscle reaction to mechanical shocks applied on the trunk in a lateral direction (102, 120, 121) but with highest responses in ES and EO (120).

**Table 3.** Electrode positions of electromyography.

<table>
<thead>
<tr>
<th>Muscles</th>
<th>Application of electrodes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper posterior neck muscles</td>
<td>At the level at vertebrae C2 between the uppermost parts of m. trapezius and m. sternocleidomastoid, i.e. over the splenius capitis</td>
</tr>
<tr>
<td>UN</td>
<td></td>
</tr>
<tr>
<td>Upper trapezius (UT)</td>
<td>Halfway between C7 and the acromion</td>
</tr>
<tr>
<td>Erector spinae Lumbar level (ES)</td>
<td>3 cm lateral to the L2-L4 spinous processes</td>
</tr>
<tr>
<td>External oblique (EO)</td>
<td>15 cm lateral of the umbilicus</td>
</tr>
</tbody>
</table>

**Table 4.** Maximum voluntary contraction (MVC) tests. Isometric MVCs for each muscle performed twice during 5 seconds in static conditions using manual or rigid resistance. Inspired by: Peter Konrad. The ABC of EMG. A practical introduction to Kinesiological Electromyography (122).

- **Upper neck (UN) Splenius capite First test:** Seated on a flat chair with the head in neutral position with a headband making rigid resistance. The participants could brace against manual support during maximum lateral flexion.

- **Upper neck (UN) Splenius capite Second test:** Seated on a flat chair with the head in neutral position with a headband making rigid resistance. The participants could brace against manual support during maximum extension.

- **Trapezius (UT):** Erect seated with an arm sling over the shoulder, height adjusted to give rigid vertical resistance at instant shoulder elevation.

- **Erector spinae (ES):** Lie prone on a bench, hands held beside the head, with legs and pelvis buckled to the bench by a belt. Do a back extension and at horizontal level add manual resistance.

- **Obliquus externus abdominis (EO):** A side laying position with leg and hip fixation. Subject flex up and at horizontal level add manual resistance.
EMG root mean square (RMS) amplitude where normalised to isometric maximum voluntary contractions (MVC). The MVCs were performed twice in static conditions using manual or rigid resistance for five seconds in accordance with Table 4. The MVC, e.g. the maximum RMS amplitude during a 500 ms window, was further used for normalisation of the EMG signals from the mechanical shocks (% MVC).

EMG signals were amplified 500 times and low pass-filtered at a cut-off frequency of 500 Hz before being processed in MyoResearch 1.07.63 XP™, (Noraxon USA Inc., US). The signals were high pass-filtered with a cut-off frequency set at 10 Hz. The signals were rectified and smoothed using a 25 or 100 ms RMS running window (122). Mean amplitudes were calculated during different epochs, e.g. 400 ms following the start of seat acceleration (Paper I) or several 150 ms epochs (Paper II-III). Epoch 1 (0–150 ms) and epoch 2 (150–300 ms) following the seat acceleration from the participant’s right side, e.g. first acceleration in SSMS and second acceleration in DSMS was used. A baseline, epoch 0, 150 ms prior the seat acceleration was additionally used in paper II.

For muscle onset calculation, a computer algorithm, Matlab, (R2013B, The MathWorks, Inc., USA), automatically determined the onset of EMG activity. The calculation of muscle onset was in accordance with the description by Stensdotter et al. (123). Subsequent to the automatic determination, onsets were checked manually so as to avoid obvious misplacements or artefacts (124). Trials were excluded if the onset of EMG activity occurred after more than 200 ms after the perturbation to avoid confounding between automatic and voluntary reactions (4, 125).

**Statistical analysis**

Parametric tests were used for data that were normally distributed and in cases of skew-symmetric (Paper II & III), the variables have been log-transformed. The p-value was set to 0.05 for all analysis. Analyses were performed using IBM SPSS version 20–22 (IBM Corp. Released 2011. IBM SPSS Statistics for Windows, Armonk, NY: IBM Corp.) and Excel version 2010-13 (Microsoft Inc., USA).

The averaged kinematics values and EMG amplitudes were analysed using a two way repeated measurement analysis of variance (ANOVA) (Paper I). The two within subject factors were the different seat accelerations and conditions (Cognitive task or No task). The differences between left and right side muscles amplitude was calculated with a paired t-test. Associations between EMG amplitude and neck kinematics were analysed using Pearson’s correlation coefficient.
The muscle activity and the kinematics for each repetition were analysed using linear mixed models (Paper II & III). There were three analyses of muscle activity (Paper II) and one analysis (Paper III) (Figure 1) using linear mixed models. Model 1; comparing the first five repetitions. Model 2; comparing the effect of different head postures and the order of repetition. Model 3; comparing the pre (repetition 4-5) and post (repetition 16-17) values (Paper II). In the last analysis (Model 4), the three types of DSMs and the SSMS, using repetition 3,4,5,16,17, was compared (Paper III). The fixed factors considered in the models and included depending on the question were: Side of muscles (L, R); Muscles (EO, ES and UN); Epochs (0–2); Repetitions; Posture (L, R); Pre/Post factor; Type of mechanical shock (SSMS and DSMs Fast/Medium/Slow). Various random effects where considered in the models, using a diagonal covariance structure model, and decisions were based on the factors contributing with largest variations on the dependent factor. All two-way interactions were tested, but only those significantly improving the model were included. A Bonferroni correction was implemented when multiple comparisons were made. In paper III, different covariates were tested, e.g. kinematics (ROM and acceleration), while in paper II the kinematics (angle displacement) were analysed separately using the mixed model. The residuals were analysed and concluded to be normally distributed.

The intra-class correlation coefficients was calculated using model ICC_{2,1} absolute agreement (Paper IV) (126). The mean value of 5 repetitions from each specific posture and segment was used except in the intra-subject reliability where every repetition was used separately. The correlation was assessed according to the following criteria: <0.75, Poor to moderate; 0.75–0.90, Good; 0.91-1, Adequate reliability for clinical measurement (127). The Standard Error of Measurement \( SEM = \sqrt{(\sigma^2 + \sigma^2_{\text{residual}})} \) (128) and Smallest Detectable Change \( SDC = 1.96\sqrt{2*SEM} \) (129) was calculated for the absolute reliability (agreement). The SEM, represents the measurement error while the SDC provides information about the smallest degree of detectable change between two measurements.

**Sample size**

A sample size calculation using neck kinematic data from a similar setup revealed that 19 participants were needed in paper I, effect size = 0.8 (alpha level 0.05). Four extra participants were added, since the risks were inconsiderable and technical problems were suspected to occur. There were no prior data to calculate the sample size for paper II & III. The 19 participants needed in paper I with one extra participant in case of technical problems was assumed to be a good estimation. Nine participants were needed in paper IV, power 0.8 (alpha level 0.05), for four testers when the assumed variation
between individuals was five times greater than the variation within the individual.
RESULTS

The results from paper I–III have been compiled under the subheadings kinematics and EMG. The reliability in paper IV is presented separately.

Kinematics

The angle displacements for each segment in frontal plane differentiated in direction between a SSMS and a DSMS caused by the direction of the first seat acceleration is shown in Figure 5. When the seat accelerated, the pelvis was mechanically activated, whereupon the pelvis started to tilt. This caused an angle displacement in the pelvis, positive or negative angle depending on the direction of acceleration. Due to inertia, the trunk reacted later than the pelvis causing an angle displacement with direction opposite that of the pelvis, e.g. a positive angle in the pelvis causes a negative angle in the trunk. The pattern caused by inertia was replicated for the head in relation to the trunk.

Figure 5. The angle displacements presented for one representative participant in the frontal plane (lateral flexion) for the pelvis, trunk and neck during 2 seconds following a mechanical shock. Double-sided mechanical shock; Fast (Blue), medium (Red), slow (Green); single sided mechanical shock (grey). Shadows represents the standard deviation. The first (1) and second (2) neck peak angle displacement is displayed for the single-sided (*) and the fast double-sided mechanical shock (†).
Results

For all body segments there were significant acceleration effects $p \leq 0.01$ giving a larger angle displacement for the SSMS compared the L-SSMS (Table 5). Adding a cognitive task increased the second peak angle displacement in the trunk and neck by $p < 0.05$. In addition, a higher acceleration, combined with a cognitive task increased the second peak angle displacement by $p < 0.01$, meaning that the cognitive task affects the second angle displacement in the neck more at higher accelerations. There was no significant effect of adding a cognitive task to the first angle displacements.

Table 5. The first (I) and second (II) mean peak angular displacements and standard deviations for neck, trunk and pelvis following a single-sided mechanical shock of low peak acceleration (L-SSMS) or high peak acceleration (SSMS) combined with and without the condition a cognitive task (Yes/No). The F-value and p-value for the acceleration (L-SSMS, SSMS), condition (Y/N) and interaction effects were calculated using a two way repeated measurement analysis of variance (ANOVA). The number of participants = 23.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Task (Y/N)</th>
<th>Ang. displacement (°)</th>
<th>Acceleration</th>
<th>Condition</th>
<th>Acc*Condition</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>L-SSSM</td>
<td>SSMS</td>
<td>F p</td>
<td>F p</td>
<td>F p</td>
</tr>
<tr>
<td>Neck I</td>
<td>N</td>
<td>-2.5 (1.0)</td>
<td>-6.4 (3.2)</td>
<td>95.25</td>
<td>&lt; .01</td>
</tr>
<tr>
<td></td>
<td>Y</td>
<td>-2.8 (1.0)</td>
<td>-6.7 (2.4)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Neck II</td>
<td>N</td>
<td>4.8 (2.6)</td>
<td>7.9 (4.7)</td>
<td>35.62</td>
<td>&lt; .01</td>
</tr>
<tr>
<td></td>
<td>Y</td>
<td>5.6 (2.7)</td>
<td>10.6 (5.6)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Trunk I</td>
<td>N</td>
<td>-2.0 (0.6)</td>
<td>-3.8 (0.7)</td>
<td>394.58</td>
<td>&lt; .01</td>
</tr>
<tr>
<td></td>
<td>Y</td>
<td>-2.1 (0.6)</td>
<td>-3.8 (0.9)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Trunk II</td>
<td>N</td>
<td>7.4 (1.7)</td>
<td>15.7 (4.0)</td>
<td>230.10</td>
<td>&lt; .01</td>
</tr>
<tr>
<td></td>
<td>Y</td>
<td>7.7 (2.0)</td>
<td>16.8 (3.9)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pelvis I</td>
<td>N</td>
<td>4.0 (1.0)</td>
<td>8.8 (2.1)</td>
<td>267.53</td>
<td>.01</td>
</tr>
<tr>
<td></td>
<td>Y</td>
<td>4.1 (1.4)</td>
<td>8.8 (2.6)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pelvis II</td>
<td>N</td>
<td>-6.4 (1.3)</td>
<td>-12.2 (1.9)</td>
<td>521.10</td>
<td>&lt; .01</td>
</tr>
</tbody>
</table>

There is a clear adaptation to repeated mechanical shocks, since the first neck peak angle displacement was significantly ($p < .001$) reduced by $\geq 1.6°$ after the first SSMS shown by the interaction `Seg.*Rep.` (Table 6). The ROM, i.e. the difference between the first and second peak angle displacement, varied between different mechanical shocks in pelvis ($13.6–19.5°$), trunk ($16.7–22.2°$) and neck ($9.1–19.4°$) (Table 7). The largest ROM was found in the DSMS.
Table 6. Linear mixed models showing angle deviation with the base line set to pelvis (Segment), repetition 1 and interactions Seg.*Rep., Trunk*1 and Neck*1 for Model 1 (The first analysis in Figure 1). For the next analysis, Model 3 (Figure 1), the model is based on segment and post different head postures (Pre/Post). Model 1; adaptation for the first five repetitions with head in neutral position, Model 3; pre different head postures and post different head postures. Segments were included in the model as random effect using a Diagonal covariance structure model.

<table>
<thead>
<tr>
<th></th>
<th>Model 1</th>
<th>Model 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intercept</td>
<td>.99 ± .02 (9.7)</td>
<td>.97 ± .02 (9.2)</td>
</tr>
<tr>
<td>Repetition</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>-.04 ± .02′ (-.8)</td>
<td>-</td>
</tr>
<tr>
<td>3</td>
<td>-.03 ± .02 (-.6)</td>
<td>-</td>
</tr>
<tr>
<td>4</td>
<td>-.02 ± .02 (-.5)</td>
<td>-</td>
</tr>
<tr>
<td>5</td>
<td>-.02 ± .02 (-.4)</td>
<td>-</td>
</tr>
<tr>
<td>Segment</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Trunk</td>
<td>-.33 ± .03″ (-.5)</td>
<td>-.32 ± .03″ (-.7)</td>
</tr>
<tr>
<td>Neck</td>
<td>-.14 ± .04″ (-.2)</td>
<td>-.21 ± .04″ (-.3)</td>
</tr>
<tr>
<td>Seg.*Rep.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Trunk*2</td>
<td>.03 ± .02 (.7)</td>
<td>-</td>
</tr>
<tr>
<td>Trunk*3</td>
<td>.01 ± .02 (.3)</td>
<td>-</td>
</tr>
<tr>
<td>Trunk*4</td>
<td>.01 ± .02 (.1)</td>
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</tr>
<tr>
<td>Trunk*5</td>
<td>.01 ± .02 (.2)</td>
<td>-</td>
</tr>
<tr>
<td>Neck*2</td>
<td>-.08 ± .02″ (-.6)</td>
<td>-</td>
</tr>
<tr>
<td>Neck*3</td>
<td>-.09 ± .02″ (-.7)</td>
<td>-</td>
</tr>
<tr>
<td>Neck*4</td>
<td>-.09 ± .02″ (-.7)</td>
<td>-</td>
</tr>
<tr>
<td>Neck*5</td>
<td>-.11 ± .02″ (-.2)</td>
<td>-</td>
</tr>
<tr>
<td>Pre/Post</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pre</td>
<td>-</td>
<td>.05 ± .01″ (1.1)</td>
</tr>
<tr>
<td>Seg.*Pre/Post</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Trunk*Post</td>
<td>-</td>
<td>-.03 ± .02′ (-.6)</td>
</tr>
<tr>
<td>Neck*Post</td>
<td>-</td>
<td>-.04 ± .02′ (-.8)</td>
</tr>
</tbody>
</table>

*p<0.05, ′p<0.01, ″p<0.001, ′′ not applicable

Table 7. Range of motion (ROM) for neck, trunk and pelvis caused by a single-sided mechanical shock (SSMS) or a double-sided mechanical shock (DSMS) of slow, median or fast change of acceleration directions.

<table>
<thead>
<tr>
<th>ROM (°)</th>
<th>Mean ± SD (Median)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Neck</td>
</tr>
<tr>
<td>SSMS</td>
<td>9.1 ± 6 (7.3)</td>
</tr>
<tr>
<td>Slow DSMS</td>
<td>19.4 ± 7.9 (17.6)</td>
</tr>
<tr>
<td>Medium DSMS</td>
<td>16 ± 7.2 (14.5)</td>
</tr>
<tr>
<td>Fast DSMS</td>
<td>18.6 ± 7.3 (18.1)</td>
</tr>
</tbody>
</table>
Electromyographic measurements

Amplitude
The UT was removed from further analysis since the activity was low, less than 2% MVC in all EMG-studies regardless of type of mechanical shock or epoch. There was significantly higher mean EMG amplitude for a SSMS compared to a L-SSMS 400 ms after acceleration for both left and right side muscles (Table 8). There was no effect of adding a cognitive task during this epoch. During an SSMS from the right side a paired sample t-test showed that the EMG amplitude was higher in the agonist (left) UN (t(33.6)=4.59, p<0.01) and ES (t(32.3)=3.60, p<0.01), compared the antagonist (right side) muscle; but no differences were found in the EO.

Table 8. The mean muscle amplitude and standard deviation normalised to maximum voluntary contraction (MVC) during 400 ms after acceleration for Left (L) and Right (R) side of the upper neck (UN), erector spinae (ES) and external oblique (EO). The F-value and p-value for the single-side mechanical shock of low peak acceleration (L-SSMS) and high peak acceleration (SSMS), with and without a cognitive task (Y/N) and interaction effects were calculated using a two-way repeated measurement analysis of variance (ANOVA).

<table>
<thead>
<tr>
<th>Muscle</th>
<th>N</th>
<th>Task</th>
<th>Acceleration</th>
<th>Condition</th>
<th>Acc*Cond</th>
</tr>
</thead>
<tbody>
<tr>
<td>L UN</td>
<td>20</td>
<td>N</td>
<td>2.4 (1.0)</td>
<td>5.1 (2.3)</td>
<td>66.4 &lt; .01</td>
</tr>
<tr>
<td></td>
<td>20</td>
<td>Y</td>
<td>2.7 (1.4)</td>
<td>5.2 (2.3)</td>
<td></td>
</tr>
<tr>
<td>R UN</td>
<td>20</td>
<td>N</td>
<td>1.3 (0.7)</td>
<td>2.0 (1.5)</td>
<td>5.9 .03</td>
</tr>
<tr>
<td></td>
<td>20</td>
<td>Y</td>
<td>1.3 (0.7)</td>
<td>1.9 (1.6)</td>
<td></td>
</tr>
<tr>
<td>L ES</td>
<td>21</td>
<td>N</td>
<td>3.8 (3.1)</td>
<td>5.4 (3.2)</td>
<td>27.5 &lt; .01</td>
</tr>
<tr>
<td></td>
<td>21</td>
<td>Y</td>
<td>3.5 (1.8)</td>
<td>5.2 (2.9)</td>
<td></td>
</tr>
<tr>
<td>R ES</td>
<td>20</td>
<td>N</td>
<td>1.7 (1.5)</td>
<td>2.4 (1.9)</td>
<td>7.1 .02</td>
</tr>
<tr>
<td></td>
<td>20</td>
<td>Y</td>
<td>1.5 (1.0)</td>
<td>2.9 (3.4)</td>
<td></td>
</tr>
<tr>
<td>L EO</td>
<td>19</td>
<td>N</td>
<td>2.3 (1.2)</td>
<td>5.1 (3.6)</td>
<td>12.5 &lt; .01</td>
</tr>
<tr>
<td></td>
<td>19</td>
<td>Y</td>
<td>2.0 (1.0)</td>
<td>4.7 (4.3)</td>
<td></td>
</tr>
<tr>
<td>R EO</td>
<td>21</td>
<td>N</td>
<td>2.8 (3.0)</td>
<td>5.1 (4.0)</td>
<td>28.6 &lt; .01</td>
</tr>
<tr>
<td></td>
<td>21</td>
<td>Y</td>
<td>3.0 (2.9)</td>
<td>4.5 (3.6)</td>
<td></td>
</tr>
</tbody>
</table>

EMG amplitude due to an SSMS was increasing for each epoch, and was more evident in the agonist muscles shown by the interaction side*epoch (Table 9). The mixed model including both SSMS and DSMS (Table 10) also show the highest activity in the last epoch and was more evident in the agonist muscles. There were no EMG amplitude differences at baseline between the muscle sides other than when laterally flexing the head. The EMG amplitude adapted, shown in model 1 Table 9, with a 0.2% reduced amplitude (p<0.01) after the fourth SSMS compared to the first. The adaptation continued in model 2 and 3. There was generally lower EMG amplitude in the muscles when laterally
Results

flexing the head to the left (-0.5%) compared to the right. The EMG amplitude was 0.6–1.0% (p<0.001) higher for the fast DSMS compared to all other mechanical shocks presented in the mixed model in Table 10. A Bonferroni adjusted pairwise comparison showed an increased activity in the slow DSMS (p<0.001), but no activity difference in the medium DSMS compared a SSMS. The activity in the EO was almost 9% higher (p<0.001), while the ES was not significantly different when compared to the UN. The right side muscles were generally around 3% higher than the left side. Several interactions were found, so the activity in UN varied between 3.2 and 7.9%, ES 3.2 and 7.9% and EO 10.2 and 17.2% depending on combinations of fixed factors and interactions. The highest individual activity in UN and ES was caused by an SSMS, in left side muscle at epoch 2 while the highest individual activity in EO was caused by a DSMS in right side muscles at epoch 2.

Associations for EMG amplitude between trunk and neck

Associations were calculated for an SSMS. There was a correlation between the left side muscles with regards to the EMG amplitude for EO and ES (r=0.52, p=0.03) and between EO and UN (r=-0.53, p=0.02). A high EMG amplitude in the left side EO implies a consequently lower amplitude in the UN but a higher amplitude in the ES. Furthermore, there was a correlation between the EMG amplitude for the left UN and the first neck peak angle (r=-0.65, p<0.01) and the second neck peak angle (r=0.51, p=0.02).

Muscle onset latencies

There were detectable muscle onsets in the left side UN, ES and EO in over 90% of the cases while the right side muscles only had 60% detectable onsets (Paper II). Thirty cases (3%) of the agonist (left side) muscles and 48 cases (8%) of the antagonist (right side) muscles were further excluded due to the 200 ms time constraint. The onset latencies in the different muscles and side varied between 83-128 ms depending on combinations and model (Table 8). The EO responded around 14 ms quicker (p<0.01) than the UN in model 2 and 3 but with no difference in model 1. There were differences between muscle sides in model 2 i.e. when laterally flexing the head, with 15 ms increased onset latencies in the antagonist muscles (p=0.011). There were no generally side differences in model 1 and model 3 but the antagonist ES was 22 ms slower compared to the agonist ES in model 1. There was no adaptation present during model 1 and 3, but in model 2 the onset latency was increased by 12 ms at the fifth (p<0.001) SSMS. There was a general decrease in the onset latency (-3 ms) when laterally flexing the head to the left (with the seat acceleration) compared to the right (against the seat acceleration) but no further interaction effects.
Table 9. Mixed model for RMS amplitude (% MVC) and muscle onset latencies (ms): The baseline is set to left side muscles (Side), upper neck muscle (Muscle) and for RMS epoch 0. For model 1 and 2 (see Figure 1) the baseline includes Repetition 1, for model 2 the head posture laterally flexed at right (Posture), for model 3 the two perturbations pre different head postures (Pre/Post). Two-ways interaction are included in the model when deviation is significant. Model 1; Adaptation for the first five repetitions, Model 2; Laterally flexing the head, Model 3; Pre and post different head postures. Random effects were Muscle + Epoch for RMS and Muscle + Side for muscle onset latencies using a diagonal covariance structure model. Statistics is based on the logarithmic values and the back log estimate is presented to inform about the approximately linear values.

<table>
<thead>
<tr>
<th></th>
<th>Log Estimate ± Log SE (Back log Estimate)</th>
<th></th>
<th>Log Estimate ± Log SE (Back log Estimate)</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>RMS amplitude (% MVC)</td>
<td></td>
<td>Muscle onset latencies (ms)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Model 1</td>
<td>Model 2</td>
<td>Model 3</td>
<td>Model 1</td>
</tr>
<tr>
<td>Intercept</td>
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<td>.24 ± .06</td>
<td>.02 ± .07</td>
<td>2.01 ± .02</td>
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<td>2.01 ± .02</td>
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<tr>
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<td>#</td>
<td>-.01 ± .01</td>
<td>-.24 ± .01</td>
<td>-.03 ± .01</td>
</tr>
<tr>
<td>3</td>
<td>-.01 ± .02</td>
<td>-.04 ± .01</td>
<td>-.03 ± .01</td>
<td>-.03 ± .01</td>
</tr>
<tr>
<td>4</td>
<td>-.06 ± .02</td>
<td>-.03 ± .01</td>
<td>-.01 ± .01</td>
<td>-.01 ± .01</td>
</tr>
<tr>
<td>5</td>
<td>-.07 ± .02</td>
<td>-.05 ± .01</td>
<td>-.02 ± .01</td>
<td>-.02 ± .01</td>
</tr>
<tr>
<td>Side (L)</td>
<td></td>
<td></td>
<td></td>
<td>.02 ± .03</td>
</tr>
<tr>
<td>R</td>
<td>-.14 ± .02</td>
<td>-.02 ± .03</td>
<td>-.02 ± .02</td>
<td>-.02 ± .02</td>
</tr>
<tr>
<td>Epoch (0)</td>
<td></td>
<td></td>
<td></td>
<td>.03 ± .03</td>
</tr>
<tr>
<td>1</td>
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<td>.28 ± .03</td>
<td>.31 ± .04</td>
<td>-.05 ± .01</td>
</tr>
<tr>
<td>2</td>
<td>.76 ± .05</td>
<td>.68 ± .03</td>
<td>.75 ± .04</td>
<td>-.02 ± .02</td>
</tr>
<tr>
<td>Muscle (UN)</td>
<td>ES</td>
<td>.15 ± .08</td>
<td>.09 ± .08</td>
<td>.19 ± .09</td>
</tr>
<tr>
<td></td>
<td>EO</td>
<td>-.06 ± .07</td>
<td>-.08 ± .07</td>
<td>-.07 ± .02</td>
</tr>
<tr>
<td>Pre/Post (Pre)</td>
<td>Post</td>
<td></td>
<td></td>
<td>-.05 ± .01</td>
</tr>
<tr>
<td>Posture (R)</td>
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<td>-.07 ± .03</td>
<td>-.01 ± .01</td>
</tr>
<tr>
<td>Side*Muscle</td>
<td>R*ES</td>
<td></td>
<td></td>
<td>-.07 ± .03</td>
</tr>
<tr>
<td></td>
<td>R*EO</td>
<td>-.13 ± .02</td>
<td>-.07 ± .03</td>
<td>.08 ± .03</td>
</tr>
<tr>
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<td>R1*</td>
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<td>-.16 ± .02</td>
<td>-.18 ± .03</td>
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<tr>
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<td>R2*</td>
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<td>-.34 ± .02</td>
<td>-.43 ± .03</td>
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<td>.03 ± .02</td>
<td>.01 ± .04</td>
</tr>
<tr>
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<td>-.03 ± .02</td>
<td>-.12 ± .04</td>
</tr>
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<td>EO1*</td>
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<td>.23 ± .02</td>
<td>.22 ± .04</td>
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<td>EO2*</td>
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<td>.22 ± .02</td>
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<tr>
<td>Side*Posture</td>
<td>R*L</td>
<td></td>
<td></td>
<td>-.02 ± .02</td>
</tr>
</tbody>
</table>

#more than two decimals are needed to set a value, - not applicable, ′p<.05, ″p<.01, ‴p<.001.
Table 10. Linear mixed model (Model 4 in Figure 1) for EMG amplitude (% MVC), n=4780. The baseline; left side muscles, upper neck (UN), epoch 1 and fast double-sided mechanical shock DSMS (Fast). Deviations from baseline presented for external oblique (EO), erector spinae (ES), epoch 2, DSMS medium and slow plus single-sided mechanical shock (SSMS). Significant two-ways interaction are included in the model. Random effects were Muscle + Side using a diagonal covariance structure model. Covariates included were ROM for trunk and neck. Statistics are based on the logarithmic values. The back log estimate is presented to show the approximately linear values.

<table>
<thead>
<tr>
<th>EMG amplitude (% MVC)</th>
<th>Log Estimates ± SE (Back log Estimates)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Intercept</strong></td>
<td>.62 ± .06 (4.2)</td>
</tr>
<tr>
<td><strong>Main effects</strong></td>
<td></td>
</tr>
<tr>
<td>Muscle (UN)</td>
<td></td>
</tr>
<tr>
<td>EO</td>
<td>.49 ± .07&quot; (8.7)</td>
</tr>
<tr>
<td>ES</td>
<td>.21 ± .08 (2.6)</td>
</tr>
<tr>
<td>Epoch (1)</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>.21 ± .04&quot; (2.6)</td>
</tr>
<tr>
<td>Side (L)</td>
<td></td>
</tr>
<tr>
<td>R</td>
<td>.24 ± .02&quot; (3.1)</td>
</tr>
<tr>
<td>Shock (Fast)</td>
<td></td>
</tr>
<tr>
<td>SSMS</td>
<td>-.11 ± .02&quot; (-.9)</td>
</tr>
<tr>
<td>Slow</td>
<td>-.06 ± .02&quot; (-.6)</td>
</tr>
<tr>
<td>Medium</td>
<td>-.12 ± .03&quot; (-1.0)</td>
</tr>
<tr>
<td><strong>Interactions</strong></td>
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<td>EO*R</td>
<td>.08 ± .02&quot; (.8)</td>
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<tr>
<td>ES*R</td>
<td>-.10 ± .02&quot; (-.9)</td>
</tr>
<tr>
<td>Shock*Side</td>
<td></td>
</tr>
<tr>
<td>SSMS*R</td>
<td>-.26 ± .02&quot; (-1.9)</td>
</tr>
<tr>
<td>Slow*R</td>
<td>-.15 ± .02&quot; (-1.2)</td>
</tr>
<tr>
<td>Medium*R</td>
<td>-.18 ± .02&quot; (-1.5)</td>
</tr>
</tbody>
</table>

*p<.05, "p<.01, "p<.001

**Reliability**

The inter-test reliability (Paper IV) when four testers used 5 IMUs to determine absolute angles in specific postures for five different segments in sagittal plane was poor to moderate. The variability between testers were ICC; 0.24–0.74, SEM; 4.1–11.5°, and SDC; 11.4–31.9° (Table 10). The intra-test reliability, when one tester measured the same participant 4 times, proved to be good in 9 and poor to moderate in 6 combinations of postures and segments. The variability between test occasions was ICC; 0.37–0.90, SEM; 2.9–7.1°, and SDC; 8.0–19.6°. The intra-subject reliability, e.g. the participant’s ability to adopt the same posture, proved to be adequate in all cases.
Table 11. Inter-tester (between testers), intra-tester (within one tester) and intra-subject (within participant) reliability. Absolute angles presented as means and standard deviations, intra-class correlation coefficients (ICC2,1 absolute agreement), standard error of measurement (SEM), Smallest detectable change (SDC) in a test retest situation for different postures and placement of inertial measurement units (IMUs).

<table>
<thead>
<tr>
<th>Posture</th>
<th>Seg.</th>
<th></th>
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<th></th>
<th></th>
<th></th>
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<th></th>
<th></th>
<th></th>
</tr>
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<tbody>
<tr>
<td></td>
<td>n</td>
<td>Mean±SD</td>
<td>ICC (95% CI)</td>
<td>SEM</td>
<td>SDC</td>
<td>n</td>
<td>Mean±SD</td>
<td>ICC (95% CI)</td>
<td>SEM</td>
<td>SDC</td>
<td>ICC</td>
<td>SEM</td>
</tr>
<tr>
<td>Seated Lumbar kyphosis</td>
<td>Head</td>
<td>6</td>
<td>106±15</td>
<td>.68 (.20-.92)</td>
<td>8.0</td>
<td>22.3</td>
<td>8</td>
<td>98±13</td>
<td>.81 (.55-.95)</td>
<td>5.1</td>
<td>14.1</td>
<td>.93</td>
</tr>
<tr>
<td></td>
<td>Th2</td>
<td>10</td>
<td>42±12</td>
<td>.60 (.31-.86)</td>
<td>7.6</td>
<td>21.0</td>
<td>7</td>
<td>47±9</td>
<td>.74 (.43-.94)</td>
<td>4.3</td>
<td>11.8</td>
<td>.95</td>
</tr>
<tr>
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<td>Th12</td>
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<td>88±9</td>
<td>.63 (.32-.87)</td>
<td>5.6</td>
<td>15.6</td>
<td>7</td>
<td>92±7</td>
<td>.82 (.57-.96)</td>
<td>2.9</td>
<td>8.0</td>
<td>.96</td>
</tr>
<tr>
<td></td>
<td>L3</td>
<td>10</td>
<td>99±8</td>
<td>.67 (.39-.89)</td>
<td>4.5</td>
<td>12.5</td>
<td>8</td>
<td>107±8</td>
<td>.80 (.54-.95)</td>
<td>3.8</td>
<td>10.4</td>
<td>.97</td>
</tr>
<tr>
<td></td>
<td>S2</td>
<td>10</td>
<td>104±11</td>
<td>.74 (.49-.92)</td>
<td>5.5</td>
<td>15.2</td>
<td>8</td>
<td>109±8</td>
<td>.77 (.49-.94)</td>
<td>3.9</td>
<td>10.9</td>
<td>.96</td>
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<tr>
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<td>118±12</td>
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*Significant differences between testers
Discussion

Main findings
The aim of this thesis was to describe and analyse seated postural reaction in healthy participants when exposed to laboratory-induced lateral mechanical shocks and to evaluate measurement of seated spinal postures. The main findings are that the characteristics of the mechanical shock influence the seated postural reactions, but the reactions are small. A quick adaptation exists in response to repeated mechanical shocks, and adding a cognitive task affects the postural reactions. Measuring seated spinal postures with IMUs could be performed with greater reliability if carried out by the same tester every time, but it is still insufficient to distinguish small differences in posture.

Kinematics
The movement pattern found after a mechanical shock started with a lateral flexion in the pelvis followed by a reversed trunk and neck lateral flexion confirming a caudocranial movement pattern, which has been proposed by other studies (87, 94). A larger angle displacement was expected and found in the spine during an SSMS compared to an L-SSMS, due to higher seat acceleration. The result conforms to earlier research on mechanical shocks in fore and aft directions (85).

Adding a cognitive task increased the second peak angle displacement of the neck and thorax. The increased displacements are consistent with impaired performance found at perturbations with a cognitive task in standing positions (104-106). The change is attributed the “central set” since evidence suggests that the cortex is involved in changing the automatic reaction due to alterations in cognitive state. The differences occurred mainly in the second and not in the first angle displacement, which may be explained by the reaction time. The first peak angle displacement occurred less than 200 ms after perturbation, meaning that the automatic reaction might have begun but that the effect has not yet appeared. The second peak angle occurred more than 500 ms after perturbation when voluntary reactions would also be involved which most likely also could be affected by attentional interference.

A significantly decreased angle displacement in the neck occurred after the first SSMS, which suggests a quick adaptation. Previous studies of forward perturbations have reported both increased and unchanged head angles after repeated mechanical shocks (82, 88). One possible explanation for the decreased neck angle in this thesis might be an adaptation of the postural reaction in the neck region i.e. a change in the central set. Another possible explanation is that the adaptation is caused by a change in the lower segment. The relative neck angle depends on the relation between the IMUs on the head
and TH2, but also between TH2 and L3 causing the trunk angle. If the change were only to occur in the neck, we have to assume that the trunk angle is unchanged which is not totally secured. The ROM, i.e. the difference between the first and second peak angle displacement, in the trunk and neck increased in a DSMS compared to a SSMS. The ROM in the neck was, however, less than 20°, which corresponds to only about 25% of the normal capacity (130) and within a neutral zone rather than at the end range of a lateral flexion. Based on the results from the kinematic measurements showing small movements, this region of the spine should be at limited risk of compression/stretching neural elements or abnormal deformation of ligaments, joint capsules, annular fibres that may produce pain sensation and MSDs (75).

Electromyographic findings

The mean amplitudes in the UT were less than 2% MVC in all studies, which is higher than the reported mean amplitude measured among forestry drivers in field conditions (131), but less than reported mean amplitude in bus drivers (132). Periods of trapezius activity exceeding 2% MVC, an indicator of active periods involving MU of low thresholds, have previously been identified during work, e.g. computer workers, secretaries, health care in an attempt to find associations between pain in the shoulder/neck region and activity in the trapezius muscle (133). Since the UT did not exceed the 2% MVC limit in the studies of this thesis, it was excluded from further analyses. The lack of reaction in the UT is in line with results by Sacher et al. (92) that only used a SSMS of low peak acceleration with hands resting at the legs. The trapezius muscle has commonly been associated with pain in the neck and shoulder (118, 119), suggested to be caused by e.g. forceful exertions, high level of static contractions, prolonged static loads and extreme postures. However, according to this result, the upper part of trapezius appears not to contribute to stabilisation of the spine during mechanical shocks in lateral directions. Higher EMG amplitude was expected and found in the UN, ES and EO during a SSMS compared to a L-SSMS, due to a higher seat acceleration, which is consistent with earlier research of mechanical shocks in other directions (85). Introducing a cognitive task did not affect the EMG amplitude in the same way as occurred within the kinematics. This may be attributed to the fact that the changes in the kinematics occurred at a later period of time, or that the period analysed was too long to detect small changes. Unpublished analyses with epochs down to 100 ms did not change the outcome, which is why the 400 ms epoch was still considered valid. An analysis including even shorter epochs, especially in the time window for the automatic reactions, might have found amplitude changes due to alteration of the central set. However, since the EMG amplitude seems small it was considered to be of limited interest to analyse it any further. The postural reaction following an SSMS from the
A participant’s right side showed more activation in the left UN and ES (agonist side) compared to the right side, i.e. reciprocally activated muscles which suggest a sloppy strategy. The activity in EOs on the other hand was more equally distributed i.e. a co-contraction was exhibited, suggesting a stiff strategy. A mix of reciprocal and co-contraction activation patterns rather than a pure strategy in response to an SSMS has been suggested previously, and the studies in this thesis are in agreement with these results (94, 134).

The EMG amplitudes were significantly decreased after the third SSMS when the participants held their heads in a neutral position. The order of repetitions continued to reduce the amplitude at later repetitions. A decreased amplitude conforms to the results of Blouin et al. (88) and Siegmund et al. (82), who found similar results in the sternocleidomastoid, scalenes and trapezius following repeated forward perturbations. The CNS, i.e. the central set, seems to prefer and adjust the postural reaction to a more relaxed strategy, but with maintained postural stability and orientation. The estimated EMG amplitudes with a laterally flexed head seem higher compared to a neutral head posture. This is contradictory to Kumar et al. (97) reporting reduced EMG amplitudes, e.g. splenius capitis, and trapezi, for a participant seated with a rotated neck compared with a neutral head posture. The different outcomes may be explained by biomechanical differences between a rotated and a laterally flexed neck.

The results in this thesis show that the EMG amplitudes after the fast DSMS were significantly higher compared to the other shocks. Complex shocks is rarely studied, but Xia et al. (99) found increased back muscle activity in a complex shock going in one direction followed by a new direction compared to a SSMS. This show that complex shocks, which are probably more realistic when driving on irregular terrain compared to the SSMS, are more demanding. Nevertheless, the single highest estimated mean amplitude in UN and ES, approximately 8% of an MVC, seem to occur due to an SSMS. One confounder to this result could be the effect of reduced amplitude caused by adaptation during repeated SSMSs before the test of the DSMS. The mean amplitude in the UN seems similar to reported levels in static work conditions with awkward head postures (135), but now experienced for less than a second. The ES activities are at the same level as in UN, which is low considered that ES is a postural muscle. The activity in EO during a DSMS exceeds 10% regardless of side, shock or epoch, which was significantly higher compared to the other muscles. The high activity implies that the EO is active in restoring the lost equilibrium caused by the lateral mechanical shock and that the activity increases when introducing DSMS that disturbs the equilibrium more.
The right side muscles are opposite the direction of the first DSMS seat acceleration, which is why they should be stretched first and are suggested to react first (94, 96). This matches to our results, since a general 3% higher activity in the right side muscles was found which was most evident for the fast DSMS. The second acceleration in the fast DSMS occurred 204 ms after the first acceleration, which implies that automatic reactions (≤180 ms after acceleration onset) (4) due to the fist acceleration had already started. The general activity was increased in epoch 2 compared to epoch 1 but in the interaction Epoch*Side the activity decreased in the right side muscles. This may be explained by an increase in the left side muscles responding to the negative acceleration and a decrease in the right side muscles, since the first reaction reduces in activity. The second reaction (i.e. due to the second, negative acceleration) therefore occurred while the first reaction was still active, causing possible co-contractions.

**Associations for EMG amplitude between trunk and neck**

When analysing the SSMS there was a negative correlation between the muscles in the agonist side with regard to the EMG amplitude for EO and UN i.e. a high activity in the EO reduces the activity in the UN. Since the amplitude in the UN due to a DSMS seems not to be increased compared to a SSMS but the EO are significantly increased, the association seems to be correct even for a DSMS. This suggests that the EO is important when restoring equilibrium after a mechanical shock in seated positions.

**Muscle onset latencies**

Side differences in muscle onset latencies were found only in ES when the first five SSMS were analysed. With the head laterally flexed, the onset latency was as expected generally longer in the antagonist muscles (right side). When the UN, ES and EO were compared, no differences were found in the first model, while the EO was a quicker responder compared to the UN in model 2 and 3 confirms a caudio-cranial muscle onset pattern. Longer onset latency in the antagonist compared to the agonist splenius capitis has previously been shown. The muscle onset latency in the agonist side complies with our estimated onset latency in UN, while the reported time in the antagonist side was much longer (96). The trend is longer onset latencies in the antagonist muscles but this cannot be statistically determined. The small reactions in the antagonist ES and EO reduced the number of determined onsets that could probably affect the results. A mix of reciprocal and co-contractions activation is in line with earlier suggested responses (94, 134).

No adaptation was found for the muscle onset latencies when the head was in a neutral position. This is in agreement with Siegmund et al., who tested sternocleidomastoid and cervical paraspinal muscles (82). In contrast to those results, we observed an adaptation effect causing increased muscle onset
latency when the participants flexed their heads. Prestretching the muscles before being exposed to a mechanical shock may initially reduce the muscle onset latency slightly compared to a neutral muscle. However, due to increased muscle onset latency in later repetitions for a prestretched muscle, the neutral muscle will become the quicker responder over time.

**Reliability**

The intra-tester reliability using IMUs to determine absolute angles for five different segments in seated spinal postures was superior to the inter-tester reliability and mostly considered good (127). Variability within the participants’ test retest situation was rather low shown by the intra-subject reliability that was adequate in all cases. There are several causes for the lower intra-tester reliability compared to the intra-subject reliability, such as the test-leader’s ability to palpate the spinal processes, attachment of the units, and the movements of the Velcro straps and the vest on skin and clothes. For one tester measuring repeatedly the SEMs was below 5° for the majority of segments, a limit that can be of clinical importance (136). Still, in the majority of segments there has to be more than 10° differences between two measures before a clinical difference is found. The SEMs and SDCs are considered too high to detect small changes. During real life working conditions the participants would have the possibility of using backrests. Therefore, it seems necessary to use small sensors or to find alternative IMU placements in future field measurements. Also, fewer but representative IMUs, attached to the skin, would be preferred to improve the measurements in combination with ROM or relative angles as a complement to the calculated absolute angles used in this study.

**Postural reactions and physical loads**

One assumed injury mechanism related to mechanical shocks are that passive muscles or delayed muscles reactions may cause excessive movements and thereby loads and strains on passive structures (75). The results from these studies showed that none of the muscles investigated were completely passive, even though UT was only activated up to 2% MVC. Nor do the results indicate that the muscle onset latencies were delayed by the experimentally induced situations, since they were within 90 and 130 ms which is well within the automatic reaction times (4). Since the movements in the different body segments caused by the mechanical shocks were within neutral zones, the muscle reactions seem adequate to retain postural control. The postural reactions seem not to cause excessive load or strains on passive structures nor stretching muscles.

It was hypothesized that a DSMS could increase the loads on the spine by creating higher muscle activities, but this was not seen. The only muscle that
certainly increased its activity during the DSMS was the EO, even though less than 20% of its potential was used. However, both sides of the EO were active during the postural reaction causing a co-contraction, or “stiff strategy” and seems to fulfil its purpose to transiently stabilize the spine.

The modest muscle activity combined with the small movement suggest either that no more muscle activity is needed or that deeper local muscles or passive structures are involved in stabilisation of the spine during seated postural reactions caused by external mechanical shocks (90, 94). Still, the modest muscle activity can be of importance. Sustained muscle activity, especially of type I MUs, may be a primary cause of MSDs (118). Suggested consequences of sustained muscle activity are e.g. accumulation of Ca\textsuperscript{2+} in the MU and homeostatic disturbance due to limitations in the local blood supply (118).

Something that was not addressed in this thesis was the contribution of cyclic loading caused if the mechanical shocks were to be repeated frequently on a daily basis. Cyclic loading from WBV combined with the posture has been suggested as being an important mechanical factor to consider when assessing the risk of injury to the vertebral joints (137). Also, significant negative effects of WBV has been suggested, such as biochemical changes in the discs that may cause neck pain (138). Speculatively, the transient mechanical shocks might not separately cause excessive loads but might contribute to the cyclic loading that in the long term could increase the risk for pain and injury in spinal structures.

**Risk assessment and prevention for drivers**

The result from these laboratory studies suggest that exposure to a single or some 20 repeated mechanical shocks with acceleration peaks up to 14 m/s\textsuperscript{2} do not seem to cause any severe seated postural reactions. Assuming that these postural reactions are equivalent to those situations that occur during actual driving on irregular terrain, there is no support for adjusting current work risk assessments. These results therefore indicate that the passive models, without contribution of muscle activity or kinematics, used in the standards for work risk assessment may still be representative for describing the vibration exposure situation.

Accurate description of the seated spinal posture while driving may be important, but seems difficult to determine even when advanced technical equipment is used. Posture has been suggested to be an important mechanical factor to consider when assessing the risk of injury from cyclic loading on the vertebral joints. A flexed posture has been associated with a greater height loss compared to neutral or extended postures (137). A simpler model with more
rough estimates of postural reactions were chosen in paper I–III as a conclusion of the insufficient output from the more detailed model used in the fourth paper. Such a model may be more appropriate and feasible to include in future assessments. The rather modest movements, at least in the neck, indicate that a poor posture (119) rather than the actual reaction may be of importance for the development of neck pain.

The EOs seem, according to these studies, to be the most active muscles in restoring equilibrium after a mechanical shock and important for control of spinal stability. Stability training, in order to improve neuromuscular control and the strength of the EOs could, therefore, be a possible recommendation. From other studies, training regimes provide strong evidence of beneficial effects on back pain, but there are few studies that clearly show a preventive effect (139). Even though the neck muscle activity is modest, training might be beneficial in reducing the prevalence in professional drivers of MSDs in shoulder and neck reported. A supervised neck/shoulder exercise regimen, with progression towards load-situated exercises and synergy endurance-strength exercises, has previously been reported to be effective in reducing the prevalence of neck pain in helicopter pilots (140). Since their work environment has similarities to that of the professional drivers with forces acting on the neck, it would be interesting to evaluate training also for drivers of earth-moving machines.

Adding a cognitive task, i.e. an attentional demanding task, seems according to these result to increase the neck angle displacement caused by a mechanical shock. Even though the neck angle displacement increased, it was still small. However, a mechanical shock of higher magnitude caused when driving on rough terrain might cause larger displacements. One recommendations could, therefore, be to avoid other attentional demanding tasks when driving on irregular terrain.

**Methodological considerations**

The results of these studies are based on inexperienced young male participants without ongoing problems in the back and neck. Young participants were chosen so as to minimize the occurrence of rigidity of the spine, and males were chosen since the large majority of drivers are men and reduce variability. The results mimic a healthy reaction from a seated participant on an un-damped seat without previous experience of driving on terrain other than during leisure time. Our results support a quick adaptation, which was well considered in the design since practice and mean values were used to reduce the risk of adaptation in the first study and the adaptation proven in the second study. Using more experienced drivers would certainly
affect the results because of their experience in seated mechanical shocks. Since awareness has proved to influence the results, the participants were prepared that a perturbation would come, but the exact time was randomized. The seating instructions were designed to reduce compensation strategies from other parts of the body. Anthropometry might differ between professional drivers and the young tested participants, which might affect the results. However unpublished results from paper II show that anthropometric measurements do not appear to have any bearing on how much motion is eliminated in the spine in young healthy men exposed to mechanical shocks.

One consideration was that the motion simulator did not manage to deliver the accurate mechanical shocks that were expected, especially in the DSMS. The actual accelerations were deviating somewhat, especially the negative (second) acceleration at medium and slow DSMS. The discrepancy, partly explained by limited power in the motion platform, could have influenced the results. A strength was that the negative seat accelerations were tested as a covariate in the mixed model, but the covariate was excluded since it was without significant effect. The reduced accuracy despite, the strength of using a DSMS were that it provides a better estimation of the real impacts that professional drivers are experiencing, something that has been a limitation in previous studies of seated postural reactions.

A second consideration was the dropout of detectable muscle onsets. The dropouts may have several explanations such as tissue between electrodes and the source heavily filtering the signal, signal noise, artefacts from cables, small muscle reactions relative the baseline or a combination of those. For the antagonist (right side) muscles, the missing onsets were of importance since only 55% of the cases had a clear muscle onset below 200 ms and even less in the ES and EO. The ES is activated in sitting, so the lack of clear muscle onset in the right ES may be caused of a small discrepancy between reaction and the baseline level. Difficulties in finding muscle onsets in abdominal muscles in the antagonist side has been reported in an earlier study (91). A strength regarding muscle onsets was that blinded person with previous experience of EMG analyses did the algorithm calculation and the first control and adjustment. In cases of doubt regarding muscle onset, a second blinded person experienced in EMG analyses gave their opinion before the results were analysed. The muscle onset latencies might have been slightly adjusted. A sampling of 128 Hz indicates that start of acceleration could in the worst case be delayed by up to 8 ms. The onset latencies could, therefore, be slightly less than the results shown.

A further consideration was that the IMU system has not been validated according to the specific setup used in these studies. However, one problem is

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how to validate such a system, since uncertainty arises when sensors or reference markers placed on the skin or clothing move. A camera system is not necessarily flawless where the calibration is crucial, the marker setup, marker detection and model could all affect the calculations of Euler angles i.e. rotation of a body. The same is true of the sensors that also calculate Euler angles using the information from the accelerometers and gyroscopes. A comparison of such systems is, therefore, not an absolute truth. One way to circumvent this was the use of sensors with similar characteristics and the same algorithms as used in these studies being placed on a robot that performed movements with a known orientation at different speeds. A strength was that the accuracy was nearly perfect in static conditions <0.1° RMSE (Root mean square error) and with reduced accuracy during fast motion <7° RMSE in sagittal and horizontal plane (141). The increased error seems, according to their figure, to occur due to phase shifts during higher angular speeds but that peak angles or ROM still could be accurate. Altogether, this may indicate that the results in the almost static condition in paper IV may be accurate, while the precision in angle displacement in paper I–III may deviate slightly in time without really affecting the results. Similar systems has also been used with good accuracy measuring cervical movements (110, 111) and trunk posture (142).

The attachment of the IMUs and sensor size seems to be of major importance. Velcro straps and a vest were used in study IV in the belief that a future field measurement would be performed on the workers’ working clothes. This created a shield between the spine and units probably affecting the reliability. Small units are preferred, but the results strongly depend on the placements of the IMUs and are, therefore, dependent on the testers’ skill. Accurate measurements and descriptions of the spinal posture while driving seem difficult even when advanced technical equipment is used. The robustness of the IMUs used however seems to be well dimensioned since 10g has not been mentioned previously in the literature.

**Future research**

**Field studies**

A future study that would be valuable is to explore whether the modest postural reactions measured experimentally comply with authentic field measurements. During such field measurements the natural driving environment containing the terrain, seat, seated posture, driving tasks could be included. The underlying WBV with possible fatiguing mechanisms would be considered, and the real complexity of the shocks with authentic levels of acceleration would be included. Depending on the result such field studies could e.g. develop new hypotheses for testing in the laboratory or suggest
further field measurements. The result from this thesis shows that there is a large variability between individuals when it comes to seated reactions to mechanical shocks. Future field measurements together with medical examinations involving functional aspects of professional drivers should give more information on the individual capacity for sedentary work and the ability to react to mechanical shocks. Such a study could be an important step in coming closer to individually targeted training regimes and further recommendations for professional drivers.

**Arms**
The arms were pacified during the seated reactions, to focus on the trunk reactions, but may be more active and have a stabilising role during real conditions when mechanical shocks occur. A field measurement would naturally incorporate the significance of the arms.

**Deep muscles and other tissues**
The results in this thesis are based on surface EMG on superficial muscles and not on intramuscular electrodes inserted in deeper muscles. The modest muscle activity supports the suggestion that deeper muscles or passive structures are involved in stabilisation of the spine during seated postural reactions caused by external shocks (90, 94). One recent study by Olafsdottir et al. (90) used intramuscular electrodes during external single-sided perturbations in multiplier directions, and showed that different neck muscles responded with different levels of activation that seemed to be related to their mechanical advantage. Whether the reactions in deeper muscles or fatigue caused by continuous WBV in combination with more complex mechanical shocks explain the reported MSDs in the neck and back among professional drivers is still to be answered. More studies of the function of deeper muscles during mechanical shocks should be conducted.

**Seated support**
The results of postural reactions in the present thesis is all done in an erect seated posture with a plane seated surface without damping provided from the seat or the possibility to get support from a backrest. The reaction should, therefore, be more active than that done with a seat providing better support that could affect the results. However, a more supported seat does not automatically reduce the postural activity while sitting. Grooten et al. (143) showed that a more active sitting in official chairs gave less muscle activity and smaller postural movements compared to a passive sitting. If future field studies show greater postural reactions than the experimental studies, then more studies of the effectiveness of various seat supports should be carried out.
Conclusions

- Mechanical shocks of high peak acceleration cause larger postural reactions compared to a low peak acceleration.
- The angle displacement in the neck during a mechanical shock increases if a cognitive task is solved. Such activities should be avoided when driving on irregular terrains since increased movements may cause excessive loads and strains on joints and tissues.
- There is a quick adaptation to reduced neck–angle displacement and EMG-amplitudes. This proves that there is a change in the “central set” causing an improved respond readiness to mechanical shocks.
- The external obliques were the most active muscles when restoring equilibrium from mechanical shocks in lateral directions, and seem to be important for stabilising the whole spinal column. Stability training, in order to improve neuromuscular control of the EOs could, therefore, be a possible recommendation.
- Exposure to a single-sided or double-sided mechanical shock with peak accelerations up to 14 m/s² seem not to cause postural reactions to such an extent that overload of muscles or joint structures should be expected.
- Accurate descriptions of seated spinal posture seem difficult to produce even when advanced technical equipment is used. Simpler models and more rough estimates of seated postural reactions may be more appropriate in laboratory and consequently also in field measurements.
- The results show that postural control is maintained despite the small reactions in tested superficial muscles.
- Reported problems, especially in the trapezius, seem not to be a direct cause of muscle reactions due to mechanical shocks.
- On the basis of these results, there is no need to change the work risk assessment model advocated by the EU Directive.
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