SIGNIFICANCE OF INTRA-ABDOMINAL PRESSURE IN WORK RELATED TRUNK-LOADING

Morten Essendrop
2003

National Institute of Occupational Health, Denmark
Department of Physiology
ACKNOWLEDGEMENTS

I would like to thank my colleagues at the Department of Physiology at National Institute of Occupational Health (NIOH), Denmark, who have all been kind and supportive throughout the project. Thank you for a lot of Valsalva Manoeuvres.

In particular I would like to thank the following people

- My competent supervisors Bente Schibye and Michael Kjær who have always been there whenever I needed them.
- Christian Hye-Knudsen and Jørgen Skotte for their always helpful attitude and for their support in setting up equipment and conducting experiments.
- Gisela Sjøgaard, and the group of Ph.D. students at the Department of Physiology for their constructive comments and advice.
- Nis Hjortskov, Jesper Strøyer, Jesper Sandfeld, Mogens Theisen, Lars Rosendal, and the "lost boys" Thomas Bull Andersen and Henrik Næsborg. Without You it would not have been half the fun.
- Anne Faber Hansen and Klaus Klausen for valuable advice and a positive attitude.

Finally I would like to acknowledge all the participants to whom I am very grateful.
This Ph.D. project was initiated at the National Institute of Occupational Health, Copenhagen, Denmark, and is part of a major research strategy towards knowledge concerning occupational risk factors of low-back disorders among healthcare workers. Head of the department, Ph.D. Bente Schibye, National Institute of Occupational Health, Denmark and Professor MD, DMSc. Michael Kjær, Bispebjerg Hospital, provided mentorship.

The project has been approved by the ethical committee of Copenhagen and Frederiksberg, Denmark ((KF) 01-213/01 and (KF) 01-237/00). The National Institute of Occupational Health financially supported the project. Additionally part of study III has been supported by grants from “The Ministry of Culture Committee on Sports Research”.

# CONTENTS

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Summary</td>
<td>5</td>
</tr>
<tr>
<td>Summary in Danish</td>
<td>6</td>
</tr>
<tr>
<td>List of papers</td>
<td>7</td>
</tr>
<tr>
<td>List of abbreviations</td>
<td>8</td>
</tr>
<tr>
<td>Introduction</td>
<td>9</td>
</tr>
<tr>
<td>Aim</td>
<td>12</td>
</tr>
<tr>
<td>Methods</td>
<td>13</td>
</tr>
<tr>
<td>Participants</td>
<td>13</td>
</tr>
<tr>
<td>Procedure</td>
<td>14</td>
</tr>
<tr>
<td>Experimental set-up</td>
<td>14</td>
</tr>
<tr>
<td>Measurements</td>
<td>16</td>
</tr>
<tr>
<td>Statistics</td>
<td>21</td>
</tr>
<tr>
<td>Results</td>
<td>22</td>
</tr>
<tr>
<td>Discussion</td>
<td>28</td>
</tr>
<tr>
<td>Conclusions</td>
<td>32</td>
</tr>
<tr>
<td>Perspective</td>
<td>33</td>
</tr>
<tr>
<td>Reference list</td>
<td>34</td>
</tr>
<tr>
<td>Study I</td>
<td></td>
</tr>
<tr>
<td>Study II</td>
<td></td>
</tr>
<tr>
<td>Study III</td>
<td></td>
</tr>
</tbody>
</table>
SUMMARY

Introduction
Increased intra-abdominal pressure (IAP) is suggested as an active structure stabilising the spine. The aim of this thesis was to determine if IAP plays a role during trunk loading. Three hypotheses were investigated: 1) IAP increases with time as the low-back muscles fatigue when strenuous static back extension is performed. 2) IAP of a size present during normal nursing work adds to trunk stability during light sudden loads. 3) In well-trained persons, high IAP is developed sufficiently fast to enable a role in trunk stability during heavy sudden trunk loads. Three studies were performed.

Methods
In the first study, the participants were exposed to two different static back extension endurance tests. In the second study, they were exposed to standardised light sudden loadings in a quick-release set-up, while continuously holding alternating levels of IAP. Surface EMG was measured from: erector spinae muscle both lumbar and thoracic part, rectus abdominus muscle, external oblique muscle, internal oblique muscle. In the third study, well-trained judo/jujitsu fighters were exposed to heavy sudden trunk loads. They performed patient handling with an imitated patient, who fell. A 3D biomechanical model was used to calculate the biomechanical load on the participants during the patient falls. In the three studies the IAP was measured in the stomach.

Main findings and conclusion
1) Trunk extension until exhaustion initiates a strategy involving an increase in the activity of the abdominal muscles and elevation of IAP, as the back extensor muscles fatigue. The strategy seems not to relate to any specific posture.
2) IAP of a size appearing in nursing work situations seems to increase spine stiffness. The increase in stiffness is obtained through both the concomitant increase in muscle co-activation related to the generation of IAP and the IAP in itself.
3) Well-trained participants develop high IAP when the trunk is exposed to heavy sudden external loads. The development is sufficiently fast to enable IAP to function when the trunk copes with the heavy external loads.
**RESUMÉ**

**Introduktion**

Under rygbelastning menes øget intra-abdominale tryk (IAP) at kunne stabilisere hvirvelsøjlen. Formålet med denne afhandling var at bestemme om IAP kan have en stabiliserende og/eller aflastende effekt på hvirvelsøjlen under forskellige rygbelastninger. Tre hypoteser blev efterprøvet: 1) Under en vedvarende statisk rygextension øges IAP over tid i takt med at rygmsklerne udtrættes. 2) IAP, som det er målt under normalt plejearbejde, kan øge stabiliteten i truncus ved pludselige lette rygbelastninger. 3) Ved tunge pludselige rygbelastninger, udvikler veltrænede personer et højt IAP så hurtigt, at IAP har mulighed for at have en stabiliserende rolle på hvirvelsøjlen. Tre forsøg udførtes for at belyse hypoteserne.

**Metoder**

I det første studie udførte deltagerne to udholdenhedstest for rygextensioner. I det andet studie blev deltagerne udsat for lette standardiserede skub i ryggen, imens de vedvarende opholdt forskellige IAP. Elektromyografi (EMG) opsamledes i begge forsøg fra følgende muskler: toracale og lumbale del af m. erector spinae, m. rectus abdominus, obliquus externus abdominis, obliquus internus abdominis. I det tredje studie blev veltrænede judo/jujitsu udøvere udsat for tunge pludselige rygbelastninger. De udførte patientforflytninger på en imiteret patient, som pludselig lod sig falde. Den biomekaniske belastning på lænderyggen i det tidstrum hvor patientens fald forhindres blev beregnet ved hjælp af en 3D-biomekanisk model. I alle tre studier blev IAP målt i mavesækken.

**Resultater og konklusion**

1) Når en statisk rygextension holdes indtil udmattelse aktiveres bugmusklene og IAP øges efterhånden som rygextensionerne udtrættes. Den strategi synes ikke at være påvirket af kroppens stilling.

2) IAP på niveau med det der er målt under plejearbejde øger stivheden af truncus. Forøgelsen i stivhed stammer fra muskulaturen involveret i Valsalva Manøvren, der hævede trykket, samt fra IAP i sig selv.

3) De veltrænede atleter udviklede et højt IAP, når de blev udsat for den tunge pludselige rygbelastning. IAP blev opbygget så hurtigt, at det var tilstede på det tidspunkt hvor lænderygstrukturerne blev eksponeret for den store belastning.
LIST OF PAPERS

The present thesis is based on three papers. They will be referred to by their Roman numerals from the list below.


III. Essendrop M., Hye-Knudsen C., Skotte J., Hansen A.F., Schibye B. Fast development of high intra-abdominal pressure when a trained participant is exposed to heavy sudden trunk loads. 25 April 2003 accepted Spine.
## LIST OF ABBREVIATIONS

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>EMG</td>
<td>Electromyography</td>
</tr>
<tr>
<td>ES</td>
<td>Erector spinae muscle</td>
</tr>
<tr>
<td>IAP</td>
<td>Intra-abdominal pressure</td>
</tr>
<tr>
<td>IL</td>
<td>Iliocostalis lumborum muscle</td>
</tr>
<tr>
<td>ITP</td>
<td>Intra-thoracic pressure</td>
</tr>
<tr>
<td>LBD</td>
<td>Low-back disorders</td>
</tr>
<tr>
<td>LMM</td>
<td>Lumbar Motion Monitor</td>
</tr>
<tr>
<td>MP</td>
<td>Median power frequency</td>
</tr>
<tr>
<td>MVC</td>
<td>Maximal voluntary contraction force</td>
</tr>
<tr>
<td>OE</td>
<td>External oblique muscle</td>
</tr>
<tr>
<td>OI</td>
<td>Internal oblique muscle</td>
</tr>
<tr>
<td>RA</td>
<td>Rectus abdominus muscle</td>
</tr>
<tr>
<td>SL</td>
<td>Sudden load</td>
</tr>
<tr>
<td>TA</td>
<td>Transversus abdominus muscle</td>
</tr>
</tbody>
</table>
INTRODUCTION

High prevalence of low-back disorders (LBD) is found among health-care workers 16;45;56;57;77;83;85. This cannot be explained alone by the weight lifted during a workday. Even though patient handling may involve heavy lifting, the frequency of patient-handling situations is reported to be low 42;80. Epidemiological data show that the increased risk in relation to the biomechanical exposure probably relates to the specific situation of handling a patient 34;54. Patient support is shown to have the highest cause-specific injury rate per 100 full-time equivalents 75. As patients may act unexpectedly, there is a potential risk of unexpected loads during patient-handling situations. Andersen et al. 1 measured occurrence of sudden trunk movements on nurses using accelerometers, and found that on average 2.5 times during a workday high forward accelerations of the trunk occurred related to patient-handling tasks. A critical element when the low-back structures are loaded is mechanical stability 8. A stable spine can manage extremely high forces without injury; in weightlifters spine compression forces of 17,000 N have been reported 21. Active structures acting on the spine to produce stability are important. This is illustrated in vitro by the ligamentous human lumbar spine when stripped of muscle forces, buckling under the compressive load of only 90 N 28. Intrabdominal pressure (IAP) probably acts as one of the active structures adding to spinal stability when the trunk is loaded 18. Measurements of increased spinal stability caused by IAP have previously been attempted in relation to small sudden loading situations 20, and in very controlled studies with slow standardised movements with participants in side lying position on a swivel table 50. Different model studies have also shown that the IAP increases stability 19;20;38;70. These studies used levels of IAP beyond the levels measured during healthcare work 7, and loads lower than expected during a patient fall.

The transversus abdominus muscle (TA) appears to be the abdominal muscle whose activity is most consistently related to changes in the IAP 25; it can be controlled independently of the superficial abdominal muscles during postural tasks 53. Low-back patients when performing different leg and arm movements, showed a delayed activation of the TA when compared to healthy control persons 51;52. Low-back patients compared to healthy persons demonstrated a significantly different muscle response pattern during sudden loadings 78. The patients maintained co-contractions longer and had delayed onset and delayed switch off of trunk muscles compared to the healthy control persons. It is unknown whether this difference is due to a predisposing factor to LBD or a compensatory mechanism. Similar differences were observed among athletes after recovery from an acute low-back injury compared to matched healthy control persons 17.
Attempts have been made to find measures of functional capacity that can predict first time occurrence of LBD\(^9\), but the results have been contradictory\(^{16}\). Biering-Sørensen\(^9\) found that high endurance of low-back muscles seems to prevent first time occurrence of LBD. He used the Sorensen-test. The result of the test is an endurance time and is used as a direct measure of the endurance capacity in the low-back muscles. He did not measure the endurance of the abdominal muscles. Studies to find predictors of development of LBD have often concentrated on back muscles alone, and have not considered the interaction between the different low-back structures. The reason could be that the interaction is difficult to measure in a longitudinal design, and we still do not fully understand all the interactions between the different structures. However, as stated in study I, EMG measurements from the back muscles during sustained static contraction appear to deviate from EMG activity when measured on other muscles. Normally, the EMG amplitude increases towards the end of the test, but in the back muscles, it has been found that the amplitude remains constant\(^{15}\) or even decreases\(^{59}\). This indicates that low-back muscles interact with other structures during an endurance situation. In two longitudinal studies low isokinetic extension-flexion strength ratio was associated with future back injury\(^{62,84}\). Additionally, it has been suggested from a cross-sectional study that the proportion of endurance in trunk extensor muscles versus flexor muscles maybe even more important than endurance in trunk extensors alone\(^{69}\). IAP has been measured during endurance tests involving back muscles; the results were inconsistent\(^{60}\).

Other functions of increased IAP, beside increased stability, are possible; this has been studied since 1911\(^{33}\). Initially, the research was mainly concentrated on heavy lifting situations. Bartelink\(^6\) in 1957 found that the size of IAP during a lift was proportional to the lifted weight. He, and others, suggest that IAP could have a relieving effect on the lumbar spine due to an elongation of the spine and by producing a trunk extension torque unloading the erector spinae muscle (ES)\(^6,32\). The theory has been confirmed by model studies\(^{29}\). The lumbar spine compression should thus decrease, but this is not confirmed by direct measurements of intradiscal pressure\(^{3,72,92}\). McGill and Norman\(^70\) have further questioned the relieving effect of IAP. They argue that the relieving effect would be counteracted by abdominal muscle activity, especially by the rectus abdominus muscle (RA). Whether abdominal muscles can develop IAP without causing a significant flexion torque is the key point in the discussion of the role of IAP as a spine reliever. Cholewicki et al.\(^{18}\) found that it was impossible to generate IAP without trunk muscle co-contraction including ES and RA. Their measurement focused on a steady state situation, such as a lift or a Valsalva manoeuvre;
other studies investigating IAP during different lifts have not documented any relieving effect 5,46-49. Common to all these studies is that the lifts were slow and controlled, without any awkward body postures or disturbances. One study showed (7 in 10 participants) that pre-lifting co-activation of diaphragm and abdominal muscles reduce ES activity at lift-off and during the trunk erecting movement 91. In relation to a transient state, such as a sudden load or rapid limb movements the IAP development is task specific and can be generated without activity in the RA 23,25,27. A role as a spine reliever does not contradict IAP as a spine stabiliser.

If IAP relieves the spine, it would be beneficial to develop IAP during static trunk loading. We hypothesise that IAP increases with time as the low-back muscles fatigue, when strenuous static back extension is performed. Previous studies measuring increase in spinal stability caused by IAP used levels of IAP beyond the levels measured during healthcare work 7, and loads lower than expected during a patient fall. We hypothesise that IAP of a size present during normal work adds to trunk stability during light sudden loadings. As previous studies with sudden loadings exposed participants to small perturbations to avoid injury, we do not know if the human body can produce high IAP sufficiently fast to enable a function as spine stabilizer during a heavy sudden load, as might appear during patient handling.

If IAP plays a significant role as spinal stabiliser during trunk loading, both the timing between movement and IAP and the size of the IAP are essential. We hypothesise that in healthy well-trained persons high IAP is developed sufficiently fast to enable a function in trunk stability during heavy sudden loads.
AIM
The overall aim of the project was to determine whether intra-abdominal pressure (IAP) had a role during trunk loading. Specifically the aim of the project was tripartite.

- To determine if IAP increased with time as the back muscles fatigued, when continuous static loading was applied to the low-back.
- To elucidate if IAP was capable of increasing trunk stiffness, when light sudden loads were applied to the trunk.
- To see if high IAP was developed sufficiently fast to enable a function in trunk stability, when well-trained participants were exposed to heavy unexpected loading during patient handling.
METHODS

PARTICIPANTS

Participants were healthy people without previous or present back problems. The participants in study I also participated in study II. For study III, elite-trained Judo/Jujitsu fighters were recruited. Eight fighters were fighting at the Danish National Team, and two in the best National League. The general exclusion criteria were the same for all participants. The participants were excluded in the presence of either excessive blood pressure, pregnancy, angina pectoris, previous discusprolaps or the use of heart/lung medicine. Furthermore participants were excluded if pain in the back or the neck was experienced on the day of the tests. Of the entire group of participants none were excluded. Participant data are presented in Table 1.

<table>
<thead>
<tr>
<th>Study</th>
<th>I + II</th>
<th>III</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sex</td>
<td>5 women</td>
<td>5 women</td>
</tr>
<tr>
<td></td>
<td>4 men</td>
<td>5 men</td>
</tr>
<tr>
<td>Age (years)</td>
<td>36 (28-43)</td>
<td>25 (18-36)</td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.72 (1.54-1.80)</td>
<td>1.73 (1.63-1.96)</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>68 (49-88)</td>
<td>72 (58-105)</td>
</tr>
<tr>
<td>Maximal IAP (mmHg)</td>
<td>210 (127-337)</td>
<td>241 (163-331)</td>
</tr>
</tbody>
</table>

As seen from Table 1 the participant group consisted of both males and females even though the aim was not to compare between genders. In working life in Denmark, there is no regulation of the physical demands between the gender. As known, there are differences between the genders and since e.g. males are generally stronger than females ⁴, and since females at a given relative force level have higher back muscle endurance than men ⁵, the mixed genders will expand the range of the sampled data.
PROCEDURE

The basic procedure was the same for all studies. After instruction to the study, Electromyography (EMG) electrodes were placed along with additional equipment (markers study III, Lumbar Motion Monitor (LMM) study I) and the IAP catheter was inserted. The placement of the catheter was checked after insertion (see below). Reference contractions were carried out afterwards for normalisation purposes (elaborated below), along with reference IAP measurements (see study I and II). Finally the primary measurements of the study were made.

EXPERIMENTAL SET-UP

Endurance measurements

Two static back extension endurance tests were used in study I (Figure 1 A and B), hereafter referred to as upright-test (A) and incline-test (B). In the upright-test, the participant pulled with a constant force, and in the incline-test, the participant resisted a constant downward force. The tests were stopped when the participant over a 5 sec period no longer could maintain the required force or resist the constant downward force even though requests were given. The upright-test was performed in a test situation securing a constant extension torque, whereas in the incline-test, the posture was not precisely held and the external torque can change. For description of the upright-test and the incline-test, see study I. The load in the two tests was aimed to produce the same relative load on the back extensor muscles. The external torque during the incline-test was calculated from: Distance from L4/L5 to the pivotal point of the shoulder, body height and weight, centre of gravity of the trunk and the weight of body segments estimated from anthropometric tables. Since the maximal external torque a person can produce is approximately 33 % lower in upright standing compared to 45° forward inclination of the trunk, the calculated external torque during the upright-test was set to

Figure 1. Illustration of A) the upright-test B) the incline-test C) the moderated Sorensen test.
equal 66% of the calculated value in the incline-test. With a modified Sorensen test 58, we additionally tested the hypothesis from study I 36. We modified the Sorensen test making it less demanding than the original test. The trunk was held horizontal while the legs were supported on a bench inclined 10° below horizontal (Figure 1 C). Ito et al. have previously described a similar set-up 55.

Sudden applied loads
In the studies II and III, the participants were exposed to sudden loads (SL) in two different ways regarding the size of the load and uniformity. The quick-release set-up used in study II is well established at our institute and described in details previously 81. Briefly described, the set-up for creating a light but very controlled SL was constructed as a load (58N) that momentarily was attached to a wire. The wire passed a reel hereby transmitting the gravitational force on the load to a force applied perpendicular to the trunk to avoid a direct impact of the external load on the IAP. The movement of the upper body was measured by a potentiometer mounted on the reel.

With straight arms and a 30° forward inclination of the trunk in relation to vertical, the participant held a box with a weight equal to 15% of their bodyweight (7.3-13.2 kg). After the start of the data sampling nothing happened the first 5 s, but during the next 5 s the SL was trigged at a random time. Neither the participant nor the observer knew when the SL was trigged.

The participant was fixed at the hip to get isolated trunk movements, and to avoid changes related to maintenance of postural stability (Figure 2). A study made at NIOH Denmark found that this fixation was not optimal 74. Even though fixated, pelvis rotation occurred and the measured movement was a combination of movement of the spine and movement of the pelvis.

While attending the SL trials in study II, the participant was requested to continuously hold a percentage of the maximal IAP. Two trials were made in random order on each of 0%, 10%, 20%, 30% and 40% of maximal IAP. Biofeedback on IAP was given on a screen in front of the participant.
Heavier but more variable loads were applied in study III. The loads were imitated patient falls where the participant had to support the patient and prevent the fall. The so-called patient was a healthy male, present author (weight 75 kg, height 178 cm, age 33). Five different patient handling trials were performed. In two trials nothing unexpected happened, and in three trials the patient fell. In one of the fall trials, the participant knew that the patient was going to fall eventually, but the participant did not know when. We named these semi-tricked falls. In the two other fall trials, the participant did not expect the patient to fall, but he did. We named these fully tricked falls. In all situations the participant had a hold on the falling patient with both hands. The two trials without a fall served as control trials for the fall trials. Two patient handling situations served as starting point for all trials. In one situation the patient was supported from standing to sitting on a box (height 56 cm, length 66 cm, depth 48 cm) imitating a bed or a wheelchair. First time this trial was performed nothing happened, the second time the box collapsed. In the second situation the patient was moving slowly up and down in the knees, while the participant had a grip with both hands on the patient and followed the movements of the patient. Again, the first time this situation was performed nothing happened. Thereafter, the patient either fell at a non-specific moment during the movement or fell immediately.

MEASUREMENTS
Electromyography (EMG)
EMG potentials measured with surface electrodes are a sum of membrane potentials of the superficial muscle fibres. So the measured EMG is a general picture of electrical activity of
the muscle below the electrodes. The amplitude of the EMG signals can be related to force developed in the non-fatigued muscle, but there is no consensus on how to perform this relation. Rectified and low-pass filtered EMG signals in general qualitatively and positively correlate to the developed force. Larger rectified and low-pass filtered EMG signals then basically correspond to higher force in the non-fatigued muscle. For isometric contractions, the relationship between EMG signals and force is often linear at submaximal levels. However, this vary from muscle to muscle and to some extent also between participants. Also on the back muscles rectified low pass filtered EMG signals have been shown to strongly correlate to force output, but EMG signals cannot be used as a direct measure of muscle force.

As measure of the activation of relevant trunk muscles and as an indirect measure of produced muscle force, EMG was obtained from surface recordings. Pre-gelled Ag/AgCl surface electrodes (720-01-k Medicotest A/S Denmark) were used. In study I and II EMG was measured from: ES, IL, RA, OI, OE and a broad recording site intended to measure TA and oblique muscles (Electrode placement was described in study I and II). The inter-electrode distance was 3 cm except for RA (2 cm) and the broad recording site (5 cm). Surface electrodes are commonly used to measure EMG from trunk muscles. Use of surface electrodes are connected with uncertainty due to inability to record exclusively from deep muscles and by the possible "cross-talk" between adjacent muscles. For study I and II, there are limitations in the use of surface electrodes. Deep parts of the ES cannot be monitored using surface electrodes, as well as differentiation between the abdominal muscles are difficult. It would have been an advantage in order to explain the biomechanical mechanisms, if EMG activity from the TA and the deep part of the ES had been measured with wire electrodes. However, this method was not available for the present studies.

During sampling the quality of the signals were controlled on the screen of the sampling pc. The EMG signals were pre-amplified, low-pass filtered at 450 Hz and sampled with a frequency of 1000 Hz. Before additional processing of the data occurred, visual check for noise and artefacts were performed. The recorded raw data were then processed in two ways. The data were high-pass filtered with a cut-off frequency of 10 Hz, rectified and low-pass filtered with a second order single-pass Butterworth filter. To obtain information on fatigue development, a power spectrum of the EMG signals was calculated for every one-second sampling period (study I). From the power spectrum, median power frequency (MF) was calculated. To use fall in MF as a single measure of exhaustion is not beyond dispute. MF of ES is affected by both muscle length and force output and these factors were not fully
controlled in study I. We combined EMG recordings with rating of perceived physical exertion of the low-back by use of the Borg CR10 scale $^{13}$. Still it is uncertain to what extent the muscles beneath the surface electrodes were exhausted.

**Intra-abdominal pressure**

IAP was measured intra-gastrically with a pressure transducer. It was inserted through the nose, and a light local anaesthetic (Xylocaine 10 %) was offered to the participants before insertion. The required length to the stomach was estimated beforehand by a tape measure. The tape measure was guided from the tip of the nose behind the ear and downwards to the end of processus xiphoideus. The placement of the catheter was checked after insertion. In study I and II the catheter (Medtronic Microtip-catheter Medtronic inc. Skovlunde, Denmark) had two measuring sites with a mutual distance of 10 cm (diameter 3 mm.). When placed with the cranial measuring site just above the diaphragm muscle and the caudal in the stomach, the pressure curves will move in opposite directions during normal breathing. For study III a new catheter (Medtronic Single Sensor Micro Transducer Catheter Medtronic inc. Skovlunde Denmark) with only one measuring site (diameter 1 mm) was used. Before each test series the catheter was calibrated in a self-designed calibration equipment. The catheter was submerged into a flask filled with water and the wire of the catheter was going through the rubber plug in the top of the bottle. At the side of the bottle a tap connects to a tube, going to a mercury manometer. On the tube, a tap was inserted enabling inflation of air in order to raise the pressure in the closed system. Readings on the mercury manometer were then compared to the samplings from the catheter.

**Lumbar Motion Monitor**

During the incline test in study I registrations of kinematics for the low-back were made with a triaxial electro-goniometer (Chattanooga Group Lumbar Motion Monitor (LMM) Ltd. USA). At the upright test in study I, it was impossible to wear the equipment due to the already attached harness, and kinematics were therefore not measured. The LMM was attached to both the thorax and the pelvis with a semi-rigid plastic material providing two stable ‘anchors’ at the thorax and the pelvis. Thus the LMM measured the difference in inclination between the thoracic and pelvic part of the back, but gave no direct information of the curvature $^{65}$. An increased flexion, when moving from lordosis towards kyphosis increases the angle. Normal upright standing was defined as $0^\circ$. 


Force measurements
Dynamometers
For the measurement of static maximal voluntary contraction (MVC), a strain gauge ring-dynamometer (PMH electric 1999) attached to an iron chain was used. The dynamometer was calibrated against known weights before each test session. The MVC set-up has previously been described in details and evaluated for reliability.

Force plates
In study III ground reaction forces were measured by two force platforms (AMTI type OR6-5) with the participant standing with one leg on each platform or both legs on the same platform. The force plates were built into the floor, and the participants were allowed to move their feet between the platforms. During the experiment one person was assigned to observe whether the participants touched the floor or anything else with any body part below the low-back. If so the trial was discarded. This evaluation was made again during the digitisation of markers. In several trials there were minor touch between the patient's legs and the legs of the participant. In few trials this may have given a minor error in the differentiation of the force. However, in case of touch, the force curve was closely analysed along with the video recordings. When any impact or other changes in the force curves could be related to the possible touch seen on the video the trials were discarded.

Peak Motus ® Video system
In study III the trials were videotaped with a 50 Hz video system using five cameras, and digitised automatically with a Peak Motus 4.3 system. The placement of markers and the calculation from the marker to the joint centre has been described previously. All markers were digitised from at least two cameras.
3D-Biomechanical model
For calculation of the biomechanical load on the low-back, a dynamic 3D model was used. (For details on the model see 79;80). The inverse dynamic model consisted of the feet, legs, thighs, and pelvis for calculation of net torque at the L4/L5 joint. The calculation was done from the feet upwards to the L4/L5 joint. Ground reaction forces were measured with force plates, and kinematic data were obtained from video recordings. Furthermore, a previously used muscle model was used for calculation of the compression forces. The torque data on the L4/L5 joint are more valid than the calculated compression forces, since it is far more model dependent. The assumptions made for the cross-sectional muscle model were many, e.g. assumptions were made on cross-sectional area of the muscles, moment arms, number of muscles included in the model and maximal muscle force per cm². Since sudden loading situations during healthcare work do not frequently happen, cumulative load is not the main problem. Peak loads were thus investigated.

Data sampling
In all three studies, data were sampled with 1000 Hz, except for analog video data (50 Hz) and LMM data (60 Hz). In study I and II, EMG, force and IAP were sampled on a pc through an A/D-card with 16 channels. The sampling software programs were constructed in Lab View 6.0. The LMM data were sampled separately on a second pc due to incompatibility between the systems. A manual trigger secured the alignment between the systems. In study III all data were sampled by the Peak Motus pc through a 32 channels A/D-card. A time code synchronised the five video recorders and the Peak Motus pc. The signal from the IAP
catheter and the ring dynamometers were amplified (Analyser 10 PMH Electric 1999 Denmark) before the sampling on pc. An LCD screen on the amplifier was used for manual check and readings of force and IAP levels.

Data normalisation

IAP was reported either directly in mmHg or normalised to IAP obtained from maximal Valsalva Manoeuvres. Three maximal Valsalva Manoeuvres were performed separated with a few minutes pause. The highest value was used for the normalisation. In study I and II all EMG data was normalised to standardised MVC. For obtaining MVC for the back muscles, the same procedure was used in all studies. Standing in upright position, MVC was measured as described by Essendrop et al. 37. For the abdominal muscles, various methods were applied. In study I and II EMG obtained from the abdominal muscles during maximal Valsalva Manoeuvres were used for normalisation of abdominal EMG data. Since maximal Valsalva Manoeuvre will not maximally activate all abdominal muscles 25, this procedure makes comparison of relative levels of EMG between back and abdominal muscles difficult. In study III abdominal muscle MVC measurements were made in the set-up also used for the back muscles 37. Comparison with data from reference cohorts available at the NIOH was then possible.

STATISTICS

For details on the applied statistical tests, see the statistical paragraphs of the studies. Level of significance was set at P<0.05. In general the residuals were always tested for normal distribution and variance homogeneity. Parametric statistics were preferred if data were normally distributed, and when the statistical assumptions were not clearly fulfilled, it was always checked if non-parametric statistics would change the significance of the outcome. The results presented in the present studies were not affected by this problem. The software programs Minitab 13, SAS analyst V8, or SigmaStat 2.03 were used for the statistical calculations.
RESULTS

IAP development during continued static loading of the back muscles

Results on endurance time from study I along with data from modified Sorensen test \(^{36}\) are presented in Table 2. Among the three tests, the modified Sorensen test induced the heaviest load on the participants, and therefore also the shortest endurance times.

Table 2. Endurance time, external torque, and relative load for the three tests are presented. Mean values ± SD, range in brackets. *Relative load calculated as % of MVC in upright standing.

<table>
<thead>
<tr>
<th>Test</th>
<th>Endurance time (s)</th>
<th>External torque (Nm)</th>
<th>*Relative load (% MVC torque)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Modified Sorensen test (^{36})</td>
<td>270 ± 155 (130-658)</td>
<td>148±29.6 (85-182)</td>
<td>69±11.8 (51-94)</td>
</tr>
<tr>
<td>Upright-test</td>
<td>467 ± 280 (303-1112)</td>
<td>70±16.2 (47-95)</td>
<td>33±6.0 (28-47)</td>
</tr>
<tr>
<td>Incline-test</td>
<td>496 ± 355 (252-1419)</td>
<td>105±24.6 (71-143)</td>
<td>50±9.1 (42-70)</td>
</tr>
</tbody>
</table>

Figure 4. Development in IAP at different percentages of the endurance time. Results on the incline-test and the upright test from study I and results on modified Sorensen test from Essendrop \( et al. \) \(^{36}\).

Through all three endurance tests, the IAP increased progressively from the start of the test until the end (Figure 4). In the lumbar part of the ES, MF decreased significantly in all three
endurance tests. The amplitude stayed unchanged in the upright-test and the incline-test, but decreased in the modified Sorensen test. In the incline and upright-test, the EMG amplitude of the thoracic part of the ES, and of the abdominal muscles increased significantly towards the end of the test (Figure 5).

As mentioned registration of low-back kinematics was only done during the incline-test. Throughout the test, the lumbar flexion increased and changed the lumbar curvature from lordosis to kyphosis. None of the participants were able to maintain their lumbar lordosis from the start to the end of the test.
Effect and development of IAP during sudden loading

The controlled light sudden load applied in study I, was followed by a forward flexion of the trunk. In Figure 6 (A) the time needed to stop the forward flexion is plotted against the fixed levels of IAP. In Figure 6 (B) the distance of the trunk movement is plotted against IAP. Despite of a large variation there was a significant decrease in both parameters as IAP was increased.

As IAP increased we monitored the EMG activity in the back and the abdominal muscles. As expected, the EMG activity of the abdominal muscles increased along with the IAP increase (paper II), but also co-contractions were found in the back muscles (Figure 7). The EMG activity of the back muscles increased along with the IAP.

Figure 6. The mean time consumption used to stop the trunk movement (A). The mean length of the trunk movement (B). *= significantly lower than 0 % IAP.
Figure 7. Mean EMG activity for the back muscles. Number of * indicates results of the multiple comparison (p<0.05). * = significantly higher than 0% IAP, ** = significantly higher than 0 and 10% IAP, *** = significantly higher than 0, 10, and 20% IAP, **** = significantly higher than 0, 10, 20, and 30% IAP.

When exposed to heavy sudden trunk loads the well-trained participants in study III developed high IAPs. In Figure 8 maximal IAP is plotted against maximal total torque. The flexion torque was the biggest force component of the total torque, and the curve shape was similar for the total torque and the flexion torque. From the biomechanical model, compression forces were calculated. The compression force curves also looked similar to the flexion torque curves. Examples of flexion torque curves are shown in Figure 9. In Table 3 the maximal levels of torque, compression and IAP are presented for the five trials. To obtain data on timing between the IAP and the load, time gaps between the curves were calculated. In Table 4 the time gap between the IAP and the flexion torque curves is presented. If the time gap between the IAP curve and the compression force curve was presented instead, similar results occurred. In Figure 10, the time gap between the IAP curve and the compression force curve is illustrated.
Figure 8. Maximal total torque plotted against maximal IAP for the five trials.

Figure 9. IAP and flexion torque plotted against time for two associated trials. Movement up and down in the knees followed by a semi-tricked fall (A) and a fully tricked fall (B).
Table 3. Mean maximal levels of torque, compression, and intra-abdominal pressure in the trials. Significant differences between associated trials are marked in the table. * = significantly larger than movement no fall. # = significantly larger than standing to sitting no fall.

<table>
<thead>
<tr>
<th>Trial</th>
<th>Movement fully tricked fall</th>
<th>Movement semi tricked fall</th>
<th>Movement no fall</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total torque (Nm)</td>
<td>230*</td>
<td>241*</td>
<td>92</td>
</tr>
<tr>
<td>SD (Nm)</td>
<td>54.3</td>
<td>62.7</td>
<td>28.1</td>
</tr>
<tr>
<td>Flexion torque (Nm)</td>
<td>213*</td>
<td>229*</td>
<td>91</td>
</tr>
<tr>
<td>SD (Nm)</td>
<td>61.8</td>
<td>72.1</td>
<td>28.4</td>
</tr>
<tr>
<td>Lateral torque (Nm)</td>
<td>73*</td>
<td>69*</td>
<td>17</td>
</tr>
<tr>
<td>SD (Nm)</td>
<td>54.5</td>
<td>33.6</td>
<td>7.6</td>
</tr>
<tr>
<td>Rotation torque (Nm)</td>
<td>24*</td>
<td>30*</td>
<td>7</td>
</tr>
<tr>
<td>SD (Nm)</td>
<td>7.7</td>
<td>15.2</td>
<td>5.5</td>
</tr>
<tr>
<td>Compression (N)</td>
<td>5169*</td>
<td>5468*</td>
<td>2169</td>
</tr>
<tr>
<td>SD (N)</td>
<td>1127.3</td>
<td>1382.4</td>
<td>535.0</td>
</tr>
<tr>
<td>Mean IAP (mmHg)</td>
<td>130*</td>
<td>140*</td>
<td>21</td>
</tr>
<tr>
<td>SD (mmHg)</td>
<td>27.3</td>
<td>27.3</td>
<td>12.4</td>
</tr>
<tr>
<td>Number of situations</td>
<td>10</td>
<td>29</td>
<td>10</td>
</tr>
</tbody>
</table>

Table 4. Average time gap (SD) between IAP and flexion torque curves in sudden falls. If negative number IAP was developed first. *= significantly smaller than fully tricked falls.

<table>
<thead>
<tr>
<th></th>
<th>Total (n=39)</th>
<th>Fully tricked falls (n=10)</th>
<th>Semi tricked falls (n=29)</th>
</tr>
</thead>
<tbody>
<tr>
<td>90% of peak (ms)</td>
<td>-0.4 (102)</td>
<td>56 (106)</td>
<td>-21 (94.8)*</td>
</tr>
<tr>
<td>75% of peak (ms)</td>
<td>1.8 (58.5)</td>
<td>20 (58.7)</td>
<td>-5 (66.7)</td>
</tr>
</tbody>
</table>

Figure 10. Barplot of number of sudden falls divided on time gaps (at 90% of peak) between IAP and compression forces. When negative number, IAP was developed first.
DISCUSSION

Main findings

The studies reveal that abdominal muscles are activated and IAP developed during continuous static loading until exhaustion, and during both light and heavy sudden loading.

It is shown that high IAP can be developed sufficiently fast to enable a function in spinal stabilisation during heavy sudden trunk loading. The results cannot rule out that IAP may have beneficial effects during spine loading beside the effect on spinal stability.

Specific discussion on rise in IAP during sustained static back extension

The EMG activity of the abdominal muscles increased with IAP during the sustained static back extensions. An increased abdominal EMG activity during back extension is inexplicable if not aimed to produce IAP. IAP has been suggested to be a by-product of trunk muscle co-contraction; however, the results of our study suggests the opposite. As discussed in paper I there may be beneficial effects of increased IAP. If IAP produced an extension torque, this would be beneficial. However, this effect can partly be counteracted by activity in especially the RA. The low levels of IAP found in our study roughly estimated gave a 5-10% reduction of ES load if RA did not develop a corresponding flexion torque. This may be sufficient for the participant to increase the endurance. Why IAP does not increase to levels closer to the maximal IAP levels is unknown. Implication of IAP on breathing and blood circulation is possible, but no measurement can confirm this speculation. Furthermore, we know little about the isometric endurance of the TA, the diaphragm muscle, and the muscles in the pelvic floor. Their isometric endurance capacity may be a limiting factor.

As shown in study I, a change was observed in the spinal curvature during the incline-test. Spinal curvature has not previously been investigated in relation to low-back endurance. IAP has been suggested to have a direct effect on spinal curvature. The change in curvature towards kyphosis elongates the ES, which can increase the force output. In contrast to this beneficial effect, the lever arm of the ES has been shown to decrease with increasing kyphosis; meaning that the ES must produce an increased force output to maintain a certain extensor torque. Potential change in curvature in the upright-test and the Sorensen test was not investigated.

Finally, the increase in IAP could be related to spinal stability. Fatigue in the low-back muscles and a change in spinal curvature may have changed the conditions for the spinal stability.
EMG data from the back muscles suggest that the trunk load moved between different structures during the endurance test. The EMG amplitude of the lumbar part of the ES remained unchanged in the incline-test and the upright-test but decreased during the Sorensen test, which applied the heaviest load. Constant force output during static contractions would normally imply increasing amplitude with time. At the same time, the amplitude of the thoracic EMG increased, which would normally imply increased or sustained force output. Our EMG data agree with results found previously\textsuperscript{15,59}. A shift in activity between the lumbar and the thoracic part of the ES was ascribed to a change in posture and recruitment of additional parts of the ES\textsuperscript{59}. In contrast, a decline in EMG amplitude of the lumbar part of the ES was not shown by Sparto \textit{et al.}\textsuperscript{82} and Van Dieën \textit{et al.}\textsuperscript{89}; but as in study I, Sparto \textit{et al.} showed increasing EMG activity of the OE. Van Dieën \textit{et al.}\textsuperscript{90} showed that participants with high endurance have significantly more variability in EMG amplitude during intermittent isometric contractions at 70\% of MVC. They propose that alternating activity between different parts of the ES may function to postpone exhaustion of the muscle as a whole. They did not measure abdominal muscle activity.

There are limiting factors in the use of EMG as a measure of force output; furthermore, activity in deep parts of the ES is not measured with the surface electrodes. Therefore a decisive conclusion on the switch in load between different structures is difficult. Nevertheless, our results on IAP and abdominal muscle activity were similar in the three different test situations, and it is hard to explain an increased abdominal activity during back extension if not related to the measured IAP increase. The decrease in MF of the EMG for the ES indicates fatigue and with no synchronous increase in EMG amplitude, the force output has most likely decreased. As the increase in IAP occurred parallel to the assumed decrease in force output, a connection is likely. Whether the effect of the increased IAP relates to ES unloading, change in curvature, or change in stability needs is unknown.

\textbf{Specific discussion on sudden applied loads}

In light sudden loadings as they are performed in different quick release procedures, the participants only develop low (mean below 14 mmHg)\textsuperscript{35,61} or moderate IAP (mean below 64 mmHg)\textsuperscript{26} dependent on the experimental set-up. This is probably due to the very light loads that were applied in order to avoid injury of the participant. In health-care work, small levels of IAP have been measured\textsuperscript{7}, and we investigated whether these levels of IAP had significant influence on trunk performance. During the light sudden trunk loads we tried to increase spinal stiffness through an increase of IAP and succeeded. By introducing a pre-load on the
back muscles we attempted to increase IAP through the Valsalva Manoeuvre, without increasing the ES EMG activity beyond the pre-load level. Still the ES activity increased and we cannot make any definite conclusion on the isolated effect of IAP on the trunk stiffness. Nevertheless, the results of study II show that IAP of a size likely to appear in nursing work combined with the concomitant increase in muscle co-activation decreased the movements caused by the sudden load. This adds to the knowledge of IAP in relation to spinal stiffness. An increase in spinal stiffness would increase spinal stability.

In study III the judo/jujitsu athletes developed high levels of IAP, torque and compression. Higher levels of IAP have previously been measured in weightlifters. At extremely high pressures in weightlifting a decrease in left ventricular filling has led weightlifters to become dizzy and confused or even lose consciousness. None of our participants reported any symptoms during their Valsalva Manoeuvres. During the sudden loading none of the men developed more than 70% of maximal Valsalva IAP, and none of the women more than 80%, except for the woman with the lowest maximal IAP, who reached 90% of maximal IAP.

High external torques were applied to the participants when the patient fell. Compared to studies without falls our results were higher. Studies of the biomechanical load on the low-back during patient handling tasks including vertical lifts, which are most comparable to the tasks performed in study III, have reported results on compression between 4000 and 5000 N. The biomechanical model used in study III has been used to investigate the loading during patient-handling tasks as performed by ten female healthcare workers. The patient handling was performed with a patient, who had had a stroke. He had spastic paralysed muscles in both sides primarily in the left side, but had a normal function of the right arm. The extension torques and compression forces found in the 8 different actual patient handling situations ranged from $71 \pm 20$ Nm to $189 \pm 31$ Nm in torques and from $1618 \pm 322$ N to $4433 \pm 666$ N in compression, compared to $229 \pm 71$ Nm and $5468 \pm 1382$ N in study III, which was the values for the semi-tricked falls. The stroke patient weighed 88 kg, if he had fallen the external load would have exceeded the values calculated in study III. Still, the external load on the participants exposed to heavy sudden loadings was very high. For un-trained participants and indeed for low-back patients a physical load of this size would impose a risk of injury.

There are limitations concerning the application of our results to actual patient handling. Evidently the participants knew that sudden loadings would occur, consequently they were alert as they were participating in an experimental study. This implies that the trunk loads...
may be underestimated. The detrimental effect of increased IAP discussed in the literature is increased disk compression. However, for the healthy back, compression forces may imply less risk of back injury than lack of stability. Co-contractions around the spine leading to increased compression have been shown beneficial when evaluated against lack of stability. Measurements of disc compression have been made in combination with measurements of IAP, and it was shown that for the Valsalva Manoeuvre four in five measuring situations increased the spinal compression. When the participants leaned forward while holding 8 kg, increased IAP lowered the spinal compression slightly. Regarding the timing between IAP and flexion torque; high IAP was present when the low-back structures coped with the sudden heavy load. When judged visually from plots (Figure 9) the flexion torque curves were almost identical to the IAP curves at the steep increase, and the time gap between IAP and flexion torque did not significantly differ from 0. We showed a timing difference in the time to 90% of peak values, where the gap between the IAP and the flexion torque curves differed between semi and fully-tricked falls (Table 6). The peak loads were similar, but the participants were better prepared for the 'semi-tricked' falls. Supporting this, IAP tended to develop earlier than flexion torques in the 'semi-tricked' falls, and this maybe an effect of an early activation of the TA as previously measured in studies by Hodges and Richardson and Cresswell et al.

Results from Marras and Mirka suggest that IAP only increase to significant levels above 10 mmHg when more than 54 Nm of trunk torque is supported. In our results a threshold of 54 Nm is too small, but an exact level cannot be given, since we only have few observations on external torques below 100 Nm. Furthermore, the first second of the plots in Figure 9 shows that a specific flexion torque level does not determine when IAP and torque are closely timed. The link seems to be there only when the curves increase steeply. During maximal isokinetic and isometric back extensions Marras et al. 64 found that in high velocity test conditions IAP had terminated before the torque production began. There was a linear relationship between trunk velocity, and IAP-torque delay. All the lifts were voluntarily initiated, and nothing sudden therefore happened. If we had warned our participants we might have observed an even earlier development of IAP.

We know that abdominal muscle training can increase the rate of IAP development. As our participants were well-trained, the results on the timing between the torque and the IAP cannot directly be taken as an expression of that of the normal healthy worker. Additionally we do not know if the fast development of high a IAP is something the participants have learned to do through years of sports training.
CONCLUSIONS

- Trunk extension until exhaustion initiates a strategy involving an increase in the activity of the abdominal muscles and elevation of IAP as the back extensor muscles fatigue. The strategy seems to be universal and not related to any specific static posture.

- IAP of a size likely to appear in nursing work situations seems to increase spine stiffness. The increase in stiffness is obtained through both the concomitant increase in muscle co-activation related to the generation of IAP and the IAP in itself.

- Well-trained participants develop high IAP when the trunk is exposed to heavy sudden external loads. The development is sufficiently fast to enable IAP to function when the trunk copes with heavy sudden external loads.
PERSPECTIVE

The results of these studies emphasize that the functional capacity of the abdominal muscles is an important part of the low-back function in both continuous static back loading and during sudden back loading. The elevated IAP may have several beneficial effects that do not contradict one another. It cannot be ruled out that several effects act in combination. An increase in spinal stiffness and stability is probably the most profound beneficial effect. Future training interventions should consider involving abdominal muscle training in order to enable the worker or low-back patient to create IAP when needed. Bartelink \(^6\) recommends a voluntary elevation of IAP when lifting, which is in contrast to the advice from McGill \(^68\) and Marras and Mirka \(^66\) who recommend not to change the IAP voluntarily during work. In relation to the sudden loading situations, it is clear that the IAP development is far too fast to be voluntarily initiated. It is, however, probably important to ensure that the trunk muscles have the capacity to develop high IAP very fast. Strength training of the abdominal muscles and especially the TA has shown to increase the rate of pressure development but not the maximal IAP \(^24\).

In future studies we will extend our EMG measurement with intra-muscular measurements. To fully understand how IAP interacts with the trunk muscles, we need to differentiate better between the muscles. The deep parts of the ES muscle along with the deep parts of the abdominal wall are here essential. Additionally we would like to improve our possibilities of manipulating the IAP in different trunk loading situations without the use of the Valsalva Manoeuvres. It is by electrical stimulation of e.g. the diaphragm possible to produce IAP.
REFERENCE LIST


