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PhD Thesis

Henrik Koblauch

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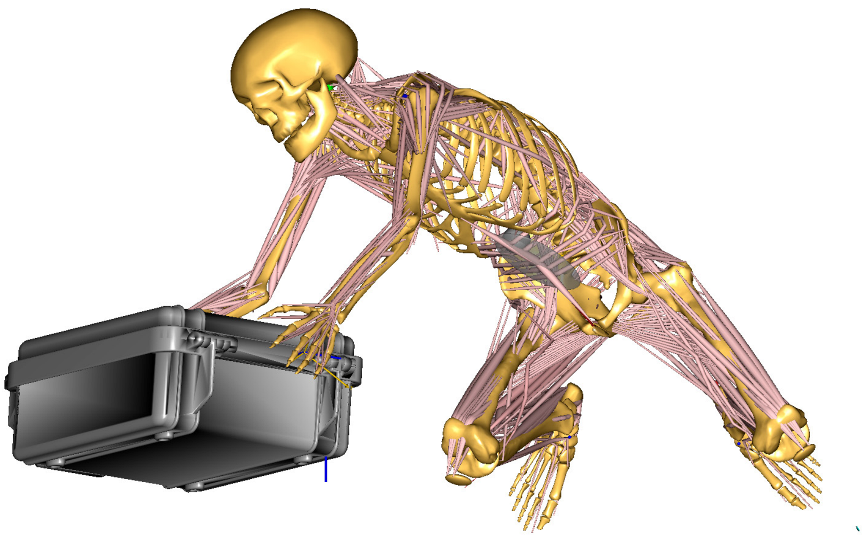
Low back load in airport baggage handlers



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Department of Neuroscience and Pharmacology
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Title: Low back load in airport baggage handlers

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1 List of Papers

This thesis is based on the following original papers, which will be referred to with their respective roman numerals. The papers can be found under the section "Papers"

Paper I

Local muscle load and low back compression forces evaluated by EMG and video recordings of airport baggage handlers.
Henrik Koblauch, Simon Falkerslev, Stine Hvid Bern, Tine Alkjær, Charlotte Brauer, Sigurd Mikkelsen, Mark de Zee, Lau C. Thygesen, and Erik B. Simonsen.
(Draft)

Paper II

The validation of a musculoskeletal model of the lumbar spine.
Henrik Koblauch, Michael Skipper Andersen, Mark de Zee, John Rasmussen, Tine Alkjær, Charlotte Brauer, Sigurd Mikkelsen, Lau C. Thygesen, Sylvain Carbes, and Erik B. Simonsen,
(Submitted to Journal of Biomechanics)

Paper III

Spinal loads in asymmetrical and dynamic lifting tasks: A modeling approach.
Henrik Koblauch, Michael Skipper Andersen, Tine Alkjær, Charlotte Brauer, Sigurd Mikkelsen, Mark de Zee, Lau C. Thygesen, and Erik B. Simonsen
(Submitted to Journal of Applied Ergonomics)

2 Abbreviations

AMS	AnyBody Modeling System
APDF	Amplitude Probability Distribution Function
DALY	Disability Adjusted Life Years
EMG	Electromyography
EMGmax	Maximal electromyographic signal
IAP	Intra-Abdominal Pressure
LBP	Low Back Pain
NIOSH	National Institute of Occupational Safety and Health
RMS	Root Mean Square
YLD	Years Lived with Disease

Contents

3 Introduction

This PhD study was an important part of the Danish Airport Cohort study. The general aims of this study were to describe and analyse the causes of musculoskeletal loading in airport baggage handlers in Copenhagen Airport. To do this a cohort of 3092 present and previous baggage handlers and a reference group consisting of 2478 men in other unskilled work without heavy lifting was established (1). The present PhD project set out to provide biomechanical input to the epidemiological exposure matrices so highly accurate measurements of the musculoskeletal loading was part of the epidemiological study.

4 Aims

- To perform a general description of the lumbar loading in the job as a baggage handler. (Paper I)

- To develop a generically useful tool to examine specific lum bar compression in a valid manner. (Paper II & III)

- To investigate the spinal loading in common work tasks for baggage handlers. (Paper III)

5 Background

5.1 Low back pain

Low back pain (LBP) is a major problem in the industrialized parts of the world. It is a massive problem for the single patient, but also a huge problem for the populations in general (3-5). Over the past two decades reports have consistently reported lifetime prevalences between 60 % and 80 % (6-9). In 15 EU countries, Norway, USA, Canada and Australia LBP is the largest burden of disease in 2010 (2;5;10) (Figure 1). LBP is the largest burden of disease measured in both Disability Adjusted Life Years (DALY) and Years Lived with Disability (YLD). DALY is defined as the number of years lost due to ill-health, disability or early death (11). YLD is years lived with disability(11). Furthermore, LBP is the sixth largest burden of disease in the world measured in DALY and the largest measured with YLD. LBP is the most activity-limiting complaint in young and middle aged and the second most frequent cause of sick-leave (12). This implies that LBP is also a large occupational health problem. Punnet et al (13) estimated that 37 % of LBP is caused by occupational exposure and many occupational groups have increased prevalence of LBP (14-21).

	Australia	Austria	Belgium	Canada	Denmark	Finland	France	Germany	Greece	Ireland	Italy
Low back pain	1	1	1	1	1	1	1	1	1	1	1
Ischemic heart disease	2	2	2	2	2	2	3	2	2	2	2
Major depressive disorder	4	3	3	3	3	3	2	3	3	3	3
COPD	9	7	8	9	4	17	14	11	11	5	14
Lung cancer	13	8	5	5	5	13	6	5	6	8	9
Alcohol use disorders	20	14	19	22	6	6	11	15	39	20	42
Stroke	15	13	10	12	7	8	12	9	4	10	7
Neck pain	6	6	7	6	8	5	8	4	7	4	5

Figure 1. Global burden of disease measured in DALY (2)

Holmstrom et al. (22) found a 1-year-prevalence of 54 % for LBP and 7 % for severe LBP in construction workers. Another occupational group with a high prevalence of musculoskeletal complaints is airport baggage handlers. Dell et al. (23) found that one in 12 baggage handlers experienced back injuries and Stålhammar et al. (24) found that more than half complained of shoulder, knee or LBP. However, these previous studies were based on limited sample sizes and there was no reference group present in either study. In a large epidemiological investigation, Bern et al. (1) found that the amount of musculoskeletal complaints increased with seniority.

5.2 The baggage handler

The baggage handlers in Copenhagen Airport are a group of only men, though primarily unskilled there are many skilled craftsmen (37 %) and a few with academic degrees (4 %). It is the primary responsibility of the baggage handler to handle baggage and make sure that baggage is correctly distributed on flights. The baggage handlers perform some different tasks but the core task is the manual handling of baggage. This implies a large amount of heavy lifting.

The average weight of a suit case is 15 kg (25) but many airlines allow baggage weights up to 32 kg (Qatar Airlines, American Airlines, British Airways etc.). When cargo is loaded on the aircraft the burdens can be even heavier. In average the baggage handler lifts 4-5 tonnes per day, and some days up to 10 tonnes (25). The baggage handling is mainly performed in three different settings: 1) Inside the baggage hall where the baggage is distributed to the correct baggage cart or container, 2) outside the narrow-bodied aircraft where the baggage is transferred from the baggage cart onto a conveyer that moves the baggage to the aircraft baggage compartment, 3) inside the aircraft baggage compartment of the narrow-bodied aircrafts where the baggage is stacked. In the baggage compartment the space is limited and the ceiling height is only about 1 m in a Boeing 737-800 (26) which is the most widely used commercial airplane worldwide. This requires the baggage handler to perform lifting in awkward positions (Figure 2) of which the most common are kneeling, stooped and sitting position. Wide-body

Figure 2. Examples of work task performed by baggage handlers in Copenhagen Airport. Top left: Baggage hall task, Top right:The conveyer-task, Bottom left: Kneeling, Bottom right: stooped positions in the baggage compartment task

aircrafts are most commonly loaded with baggage containers and the manual handling takes place in the baggage hall and not on the ramp. There is not much research available on the lifting conditions of the baggage handler. Splitstoesser et al. (27) performed a study of lifting in kneeling position and Stålhammar et al. (24) studied manual material handling in sitting, kneeling and squatting position. Furthermore, the British Health and Safety Executive have performed two studies on the risk of ill-health and how to reduce risks associated with manual handling in an airport setting (28;29). Bern et al. (1) found that 32 % of baggage handlers in the Copenhagen Airport Cohort reported complaints regarding back ache. This was significantly more than in a comparable reference group. In addition, the odds ratio for self-reported musculoskeletal symptoms increased with increasing seniority. This effect persisted when adjusted for age, BMI, smoking and leisure time physical activity. Hence, it appears that baggage handlers are at increased risk of sustaining LPB. However, this report was based on self-reported musculoskeletal complaints and not registry data.

5.3 Causes and risk factors of low back pain

Pain in the lumbar spine region may originate from many different conditions. Injured ligaments, prolapsed discs, inflammation in the facet joints, muscle spasms, compression of spinal nerve roots, vertebral periosteum are just some of the causes of pain and impairment (30). However, often no physio-pathological cause for the pain can be located and the condition is termed idiopathic. Between 14 % and 80 % of LBP are classified as "sprain and strain", "idiopathic" or "no cause" (30;31). This is probably due to lack of adequate diagnostic tools to assess injured tissue or detect a change in biomarkers. Even though idiopathic LBP has been extensively investigated, nobody has successfully located a single source for non-specific LBP.

Many risk factors for the development of LBP have been identified. High psychological work pressure (32), cigarette smoking and alcohol consumption (33), previous episodes of LBP (34), whole body vibration (35), highly repetitive work (36;37) and frequent, heavy lifting (37-45) are some of the most important risk factors for LBP. Several sub-factors, which all have a worsening effect, can be added to heavy lifting. High frequency of lifting (46), asymmetrical lifting (47), lifting in confined space (34;48), and lifting in awkward positions (34;47;48) all increase the risk of LBP. Coenen et al. (49) found that high cumulative mechanical loading of the low back estimated by observation in the workplace leads to a 2-fold increase in the risk of LBP. In general, high level of biomechanical loading is an established risk factor for LBP (9;49-53). Furthermore, Marras et al. (54) found that patients with LBP were subject to larger spinal loading than matched asymptomatic subjects due to increased activation of paraspinal muscles. In this way LBP may be a vicious circle where LBP breeds further LBP.

Another risk factor for LBP has been proposed in terms of large spinal compression and shear forces (36;52;55). These forces are increased with many of the above worsening factors. Lifting in awkward positions, lifting in confined space and asymmetrical lifting are all factors which have been shown to increase the forces on the spine (54;56-59).

So why are high compression- and shear forces damaging to the vertebrae? Van Dieën and Toussaint (60) investigated vertebral motion segment damage due to cyclic compression loading. They found that peak compression force was the leading factor in compression failure. It has been hypothesized that a possible connection between spinal loading and LBP is that high compression and shear forces can cause microfractures in the vertebral endplates and loosening of periost from the compact bone (60;61). Based on this a possible cause for non-specific LBP is microfractures with high spinal forces as the leading risk factor. However, compression and shear forces are not easily studied.

5.4 Measurement of spinal forces

It is very difficult to obtain compression and shear forces from in vivo studies. Currently, the only method for obtaining these forces directly is when a patient agrees to have an instrumented implant inserted. Spinal forces obtained by this method have been studied by a few authors (62-68), but this type of implant is extremely rare. As a consequence of this the authors have published data for public use on the orthoload-database (orthoload.com). This is extremely beneficial in many ways and especially for model validation purposes. However, many of the spinal force measurements lack kinematic descriptions of movements, which complicates the comparison with modelled estimates of spinal forces. Apart from the implant-method some authors have presented data on in vivo intra-discal pressure (69-76). However, this method is also rather inaccessible, as it is based on the insertion of a pressure gauge into the nucleus pulposus of the intervertebral disc. These measurements have been performed during different type of activities from everyday activities and body positions (68;71;73-75) to spinal manipulation (69) and heavy weight lifting (72;77). Because this level of invasiveness is preferably avoided, these data are also very rare.

5.5 Measurement of compression tolerance

There have been published several measurements of compression tolerance of spinal segments performed in vitro (78-84). In this approach a spinal segment, typically consisting of two vertebrae with the adjacent intervertebral disc, is mechanically compressed and the compression force at failure is measured. In a literature review, Jäger et al. (83) reported on a maximum compression tolerance in 776 cadaveric segments and found an average of 6180 N (SD 2660) in men and 4060 (SD 1750) in women. Furthermore, they found that the lowest compression tolerance was 1230 N and the largest was 10990 N. This large range of compression tolerances was also found by Granhed et al. (79). They found the lowest compression tolerance to be only 810 N and the largest 10090 N. In addition, Brinckmann et al. (78) found a 55 % risk of sustaining a compression injury if a segment was loaded with 40-50 % of the maximum compression tolerance 500 times. The bone mineral content in the lumbar segments is the largest predictor for the ultimate compression tolerance. A cadaver study has shown that the compression tolerance increased with 1685 N when the bone mineral content increased by one g/cm3 (79). Other factors with an influence on the compression tolerance are age, sex and nutritional status (84), which again all influence the bone mineral density.

5.6 Lifting recommendations

In an occupational setting it is unacceptable to allow workers to expose themselves to potentially damaging loads. Therefore, some recommendations for heavy lifting have been proposed (36;84-88). Some recommendations use limits of maximal compression and shear force (36;84;86), while others, like the Danish Working Environment Authority, take a more pragmatic position and recommend maximal frequency and burdens in different positions and postures (85). The National Institute of Occupational Safety and Health (NIOSH) in USA recommended a limit of 3400 N as the maximal compression force in the low back allowed during continuous manual handling. This recommendation was based on computations on a two-dimensional static model of lifting, physiological measurements and vertebral compression tolerance in cadaver studies (36). In addition to recommend limits of manual material handling the NIOSH guidelines have shown the ability to predict the risk of LBP due to lifting (89). Jäger et al. (83;84) have, based on a review of the literature, suggested another set of lifting recommendations. Unlike the NIOSH recommendations the so-called "Dortmund recommendations" are based solely on cadaver studies of vertebral compression tolerance. While the NIOSH recommendations have a fixed compression limit, the Dortmund recommendations are modulated by sex and age of the worker involved (Table 1). Based on the conclusions from the in vitro studies of compression tolerance, age and sex are imperative factors to include. However, the Dortmund recommendations completely disregard all physiological, psychological and biomechanical factors by only basing the

Age	Women	Men
20 years	4400 N	6000 N
30 years	3800 N	5000 N
40 years	3200 N	4100 N
50 years	2500 N	3200 N
≥ 60 years	1800 N	2300 N

Table 1. Dortmund recommendations (84)

recommendations on cadaver studies. Limits for shear forces during lifting have also been suggested. In a review of the literature, Gallagher & Marras (86), found that appropriate limits for shear forces were 1000 N for few (<100) cycles per day and 700 N for frequent shear loading.

5.7 Computer models

The most accessible way to estimate spinal forces is to use a computer model. Many kinds of models have been suggested including; static, dynamic, EMG-driven, hybrid, single muscle equivalent, multi-muscle, and finite element models. Since the 1980's a great variety of computer models have been published and along with increasingly powerful computers the models have increased in detail. There are advantages and shortcomings to all of them and in the following paragraphs the most important will be described.

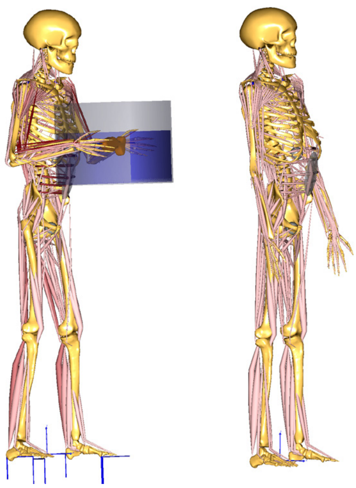


Figure 3.
Full-body models in the AMS (90)

4D Watbak (91) is a biomechanical software tool, which is easy to use. It calculates primarily the loading in the lumbar region. Watbak uses a 2D static model and single, non-wrapping joint muscle to solve the moment equilibrium. One shortcoming of the model is that it is static, so it does not account for accelerations. The model is two dimensional but it does distinguish between right and left. Furthermore, the estimation of joint moments and compressions are assumed at a single level (L4/L5) with no consideration for the equilibrium at other levels.

The AnyBody Modeling System (AMS) (92) is a commercially available software-tool for full-body musculoskeletal simulations of various activities. The main aim is to solve design problems in ergonomics, and in the AnyBody Managed Model Repository many different models for a variety of task can be found. In this system, the joint reaction forces and moments are calculated by the inverse dynamics method, where external forces and inertial properties of each segment are accounted for. The muscle redundancy issue is solved by static optimization, where different muscle recruitment criterions can be applied. A shortcoming to AMS is that it requires knowledge of the AnyScript language in which the models are programmed. Furthermore, the processing of results can be time-consuming due to the high level of detail. A similar product to this is the open source software OpenSim (93), which is slightly more user-friendly.

In finite element models it is possible to quantify the load in very complex mechanical systems. A finite element is a subdivision of a larger problem or structure. Using finite elements it is possible to estimate the load locally in the model. However, it requires an in-depth knowledge of the structure and material properties on both microscopic and macroscopic level in the different types of tissue included in the model. Previously detailed models of spinal segments and intervertebral discs have been published (94-97). Even though this method has become increasingly approachable for different occupations over the recent years, it still remains primarily an engineering tool.

For computer simulations of musculoskeletal systems a general challenge is the validity and how to verify the validity of the model (98). This is partly due to the difficulties in obtaining muscle- and joint forces from in vivo studies. Spinal models can be particularly difficult to validate, because spinal forces can only be acquired by invasive methods or from patients with instrumented implants.

In the present PhD-study we set out to investigate the lumbar load in baggage handlers. To achieve this we performed a series of EMG measurement of back and shoulder muscles, static 2D measurements of lumbar forces, and a modelling study of two common work tasks for baggage handlers, with the aim of estimating the compression and shear forces during the task. Prior to the modeling-study we performed a study of validity of the lumbar spine model in the AMS.

6 Material and Methods

6.1 Description of the baggage handling work

First, we observed baggage handlers working in the airport during a two week period and interviewed twelve of the baggage handlers about their work. Based on this information baggage handler work tasks were divided into work in the baggage hall and work on the ramp. Work in the baggage hall consisted of loading and unloading of baggage containers and belly-carts with baggage to or from a belt conveyer. A pneumatic lifting hook was available for belly-cart and open-roofed container work but could not be used with fixed-roofed containers. Work on the ramp consisted of work on the ground and work inside the airplane baggage compartments. On the ground the work was loading and unloading belly-carts with baggage to or from a belt conveyer that transported baggage between the airplane baggage compartment opening and the belly-cart on the ground. If the aircraft baggage compartment opening was low the baggage was lifted directly to or from the opening without using a conveyer. Inside the baggage compartment the work consisted of lifting the baggage to or from the ground-to-airplane conveyer and to pack or unpack the baggage inside the compartment. Some belt conveyers were extendible and flexible allowing the baggage to be conveyed to any place in the compartment (RampSnake®, Power Stow®). Depending on the size of the compartment and

The Ramp
Outside the baggage compartment
1. Loading without conveyer
2. Loading with conveyer
3. Unloading without conveyer
4. Unloading with conveyer

Inside the baggage compartment
Loading/Unloading with conveyer in
1. Standing
2. Sitting
3. Kneeling
4. Squatting
5. Stooped

Loading/Unloading with extendible conveyer in
10. Standing
11. Sitting
12. Kneeling
13. Squatting
14. Stooped

The Baggage hall
15. Loading baggage containers
16. Unloading baggage containers
17. Loading baggage-carts without lifting hook
18. Loading baggage-carts and open-roof containers with lifting hook
19. Unloading baggage-carts without lifting hook
20. Unloading baggage-carts and open-roof containers with lifting hook

Table 2.
Overview of the 20 general work tasks for baggage handlers.

conveyer belt system, loading and unloading work inside the compartment was done by one or two baggage handlers. Work positions depended on the height of the compartment relative to the height of the baggage handler and personal preferences, and were divided into standing, stooped, sitting, squatting and kneeling positions. From these basic work characteristics we defined 20 specific work tasks (Table 2).

6.2 Study designs

6.2.1 Paper I

This study was an observational study, which aimed to describe the general loading on the spine and shoulder in baggage handling work tasks. Furthermore, the aim was to investigate whether changes between three general handling tasks existed. We performed both task-based and full-day EMG measurements of back and shoulder muscles. In addition we performed 2D static load analysis on similar work tasks.

6.2.2 Paper II

This study was a validation study of the estimates of intervertebral compression forces in the spine model from the AMS. In this study we compared a series of in vivo intra discal pressure measurements in different body positions and during simple lifting tasks to the output estimates of compression forces from the AMS model in similar positions and conditions.

6.2.3 Paper III

This study was an observational study, which sought to describe the loading on the lumbar spine during common lifting tasks for baggage handlers. We recorded kinematics and kinetics by means of motion capture and used the kinematics to drive an AMS model. With the AMS model we estimated the compression and shear force, joint moments, and muscle forces.

6.3 Reduction of work tasks

6.3.1 Paper I

It was decided to collapse the 20 work tasks into 3 more general tasks: “The baggage hall”, “By the conveyor”, and “Inside the baggage compartment” for Paper I. The reduction was based on work tasks being very similar, being unmeasurable and a general question of resources. Loading and unloading at the conveyer outside the aircraft and in the baggage hall were considered to be similar. Loading and unloading with a pneumatic lifting hook were considered unmeasurable in the static computer model, as the load is carried by the hook. However, it was still a part of the baggage hall task in the EMG study, but was performed rarely, as most baggage handlers did not use the lifting hook regularly. The loading and unloading without conveyer outside the aircraft were excluded because the tasks were relatively rare, and we did not succeed in collecting sufficient data from these tasks.

After this reduction the “baggage hall” task consisted of loading and unloading belly-carts and containers, the “conveyor” task consisted of loading and unloading belly carts, and the “baggage compartment” task consisted of baggage handling in sitting, kneeling and stooped positions inside the baggage compartment. In Paper I, we did not distinguish between use of extendible conveyer in any task. For overview reasons, we report on the forces from all subtasks.

6.3.2 Paper III

In paper III we report results from two selected, very common work tasks for baggage handlers (kneeling and stooped). Furthermore, in Appendix 1 results from another 12 work tasks are reported. These 12 tasks were reduced from the original 20 tasks. The reduction was based on the same criteria as in Paper I. Both loading and unloading without conveyer were included, whereas the baggage hook tasks were not included due to modeling issues. Furthermore, the sitting tasks with and without the extendible belt loader (RampSnake®/ Power Stow®) were considered identical, because the baggage handlers, when sitting, always position a large suitcase at the end of the conveyer which the following suitcases can roll onto. Therefore, the effect is rather equal to what the extendible conveyer is used for. The baggage handlers rarely use the full functionality of the extendible conveyer and most choose not to adjust the extendible conveyer for every suitcase.

6.4 Subjects

6.4.1 Paper I

Twentythree baggage handlers, 39.6 years of age (range 24-56), were recruited for the EMG study. The first 11 subjects were selected by the nearest department leader. The remaining 12 were approached directly at the beginning of the workday and if the baggage handler agreed to participate he was included in the study. Full day EMG-measurements were obtained from the first 11 participants. In average the full day measurements lasted 4.6 (SD 1.2) hours. This was due to loss of data, mounting of equipment, termination of the workday due to injury and short shifts. The 11 full day measurements were from four baggage handlers on international ramp, two on domestic ramp, and two from the baggage hall. The task specific measurements were from seven baggage handlers on the international ramp and five from the baggage hall. There were no task specific measurements from the domestic ramp. In total the 23 participants contributed with a total of 102 task specific measurements, divided on 47 from baggage compartment, 19 measurements from the conveyer task and 36 from the baggage hall. In average the baggage hall tasks lasted (mean(SD)) 28.2 (14.0) minutes, the conveyer task 19.3 (13.0) minutes, and the baggage compartment task 22.6 (17.5) minutes

In the study of 2D static loading 10 baggage handlers were filmed in each sub task, and some were filmed in several tasks, so a total of 44 baggage handlers (40.2 years, 82.6 kg, 180.0 cm) participated. The authors recruited baggage handlers directly while they were performing the desired task. This method was mainly based on chance, and whoever performed a desired task was approached and asked to participate in the study.

Nine baggage handlers participated in both parts of the study, but this did not influence the performance in either studies.

6.4.2 Paper III

The average age and self-reported height and weight of baggage handlers in Copenhagen Airport were retrieved from Bern et al. (1) and a male subject with these average characteristics (48 years , 87 kg, 1.81 m) was recruited.

6.5 EMG measurement

6.5.1 Paper I

Bipolar EMG-electrodes (Multi Bio Sensors, Texas, USA) with a fixed interelectrode distance of 20 mm were placed on five sites on the right side: 1) m. deltoideus anterior part, 2) m. deltoideus intermediate part, 3) m. erector spinae at L4/L5-level, 4) m. erector spinae at Th12-level, and 5) descending part of m. trapezius. A reference electrode was placed on the processus spinosus of C7. Prior to electrode mounting the skin was shaved, sanded and cleaned with alcohol to reduce skin impedance. The electrodes were connected to lightweight preamplifiers equipped with an A/D-converter with 16 bit resolution. The signals were transmitted from the preamplifiers through wires to a recording box (MQ16, Marq Medical) where data were band-pass filtered (10-1000 Hz). The recording box transferred data wirelessly via Bluetooth-technology to a PC, where data was sampled using a custom-written Matlab-script. The quality of the signals was checked on the computer screen, where data were displayed in real-time. EMG was sampled at 512 Hz.

6.5.1.1. EMGmax

After the mounting of the electrodes, the maximal EMG amplitude (EMGmax) was measured during three isometric contractions for all muscles. For the anterior deltoid muscle the subject was standing with the right shoulder flexed 30 degrees. The measurement was performed while the subject pushed a tight nylon strap upwards with the back of the hand. The EMGmax recording for the intermediate deltoid was performed similarly, but with the shoulder in 30 degrees abduction. For the trapezius muscle, the subjects elevated the right shoulder against the resistance of a tight strap fixed to the floor. For both m. erector spinae parts the subjects extended the trunk against the resistance of a nylon strap around the shoulders, while the anterior part of the pelvis was supported against a plate (99).

6.5.1.2 Data processing

The full day measurements were divided into task specific measurements based on trigger signals from the start and end of tasks. Out of the total 102 we had 27 tasks specific measurements (15 baggage compartment, 12 conveyer, 5 baggage hall) from the fullday measurements. Data analysis was performed by a custom written Matlab-script. Both amplitude probability distribution functions (APDF) and rolling root mean square (RMS) amplitude were calculated. In both cases EMG-signals were band-pass filtered at 10-250 Hz using a fourth order Butterworth filter. The EMG signals were visually and manually inspected for unrealistic spikes, drift and short periods of high noise. These were rare and removed before further analysis. The method described by Jonsson et al. (100) was used to produce APDF curves. Also according to Jonsson et al. (100), three levels of activity were selected for further analysis (Figure 2). The 10th percentile (P10) was considered the static level, the 50th percentile (P50) was the median level, and the 90th percentile (P90) was considered the peak level of activity (100;101).

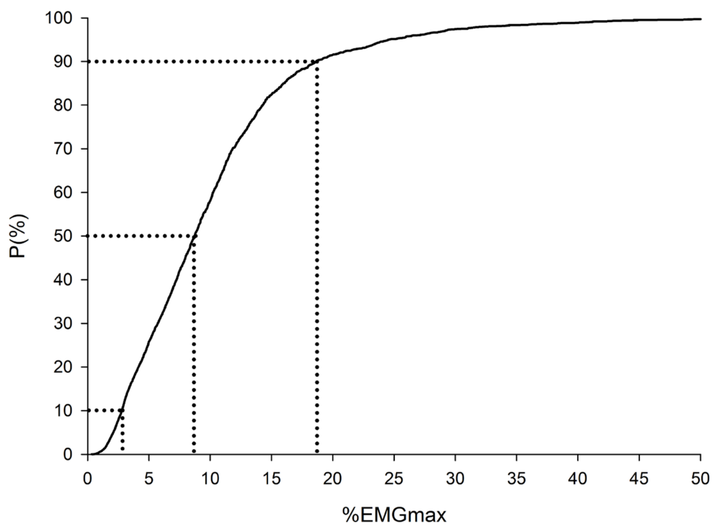


Figure 4. Example of an APDF-curve obtained from m. deltoideus intermedius. Lines show the levels p10, p50 and p90.

Rolling RMS windows of one second (RMS1), 5 seconds (RMS5), and one minute (RMS60) were calculated and expressed relative to EMGmax (%EMGmax). The peak values from the three RMS analyses along with the P10, P50 and P90 from the APDF analysis were input to the statistical analysis.

6.6 Static 2D load measurements

6.6.1 Paper I

Initially the biomechanical loading analysis was performed on all nine subtasks in the three general work tasks. However, because it was impossible to isolate the EMG measurements in the single subtasks, we decided to collapse the biomechanical loading analysis into the same three more general tasks for comparability reasons. We therefore report on the results with both methods. The compression force and flexor/extensor moment between the L4/L5 vertebrae and the right shoulder flexor moment were calculated for the same work tasks (baggage hall, baggage compartment and by the conveyor) as the EMG analysis. In each task the baggage handler was video recorded from a sagittal view. From the video five still images representing different parts of the handling task were extracted. Segment angles for foot, shank, thigh, torso, head, upper arm, forearm and hand were measured on the still images with ImageJ (National Institute of Health, USA). The segment angles were used as input to a nine segment rigid body Watbak model (University of Waterloo, Canada) which calculated the compression force and joint moment at L4/L5-level and shoulder flexor moment for the right arm.

For each of the five still pictures from every lift analysis 10 kg, 15 kg and 20 kg were used as baggage weight. To make the results comparable, all biomechanical parameters are expressed relative to body mass.

6.7 Motion capture of handling tasks

6.7.1 Paper III:

Two handling tasks were selected out of the 14 general tasks for in-depth analysis. Baggage handling in a kneeling position and in a stooped position is commonly used to handle baggage inside the air craft baggage compartment because of the limited space available. Results from the remaining models are also presented in Appendix I.

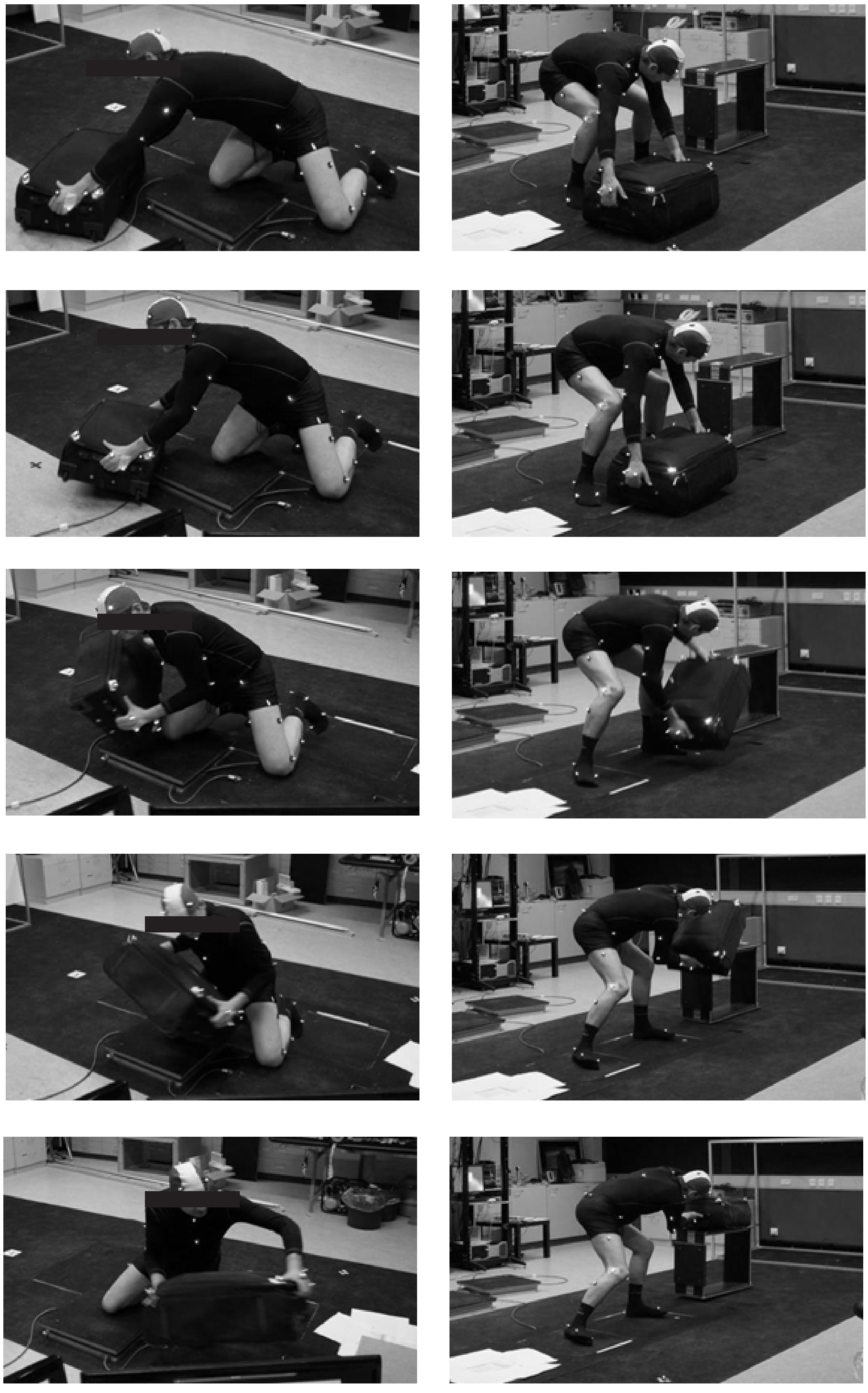


Figure 5. Time series of the two lifting tasks. Left: Kneeling. Right: Stooped

The simulation of the handling tasks took place in a lab. The setup for every task was designed based on observations of baggage handlers in Copenhagen Airport. In addition, the subject in Paper III was asked to confirm the tasks as representative before the recording.

6.7.1.1 Kneeling position

In general the subject was instructed to handle the suitcase like it was in the real airport setting. A certain speed was not specified, but a trial was considered successful if the subject approved that it was similar to lifts in the airport. The subject moved a standard suitcase (57x23.5x37 cm) from the floor using both hands and transferred it to the left and placed it on a platform 30 cm above the floor. Starting position was with the suitcase placed to the right of the subject at a 45° angle. The subject was instructed to transfer the suitcase to the designated destination at a 45° angle to the left (Figure 5). This lifting technique is frequently used by baggage handlers inside the aircraft baggage compartment lifting suitcases from the floor to a belt conveyer or vice versa.

6.7.1.2 Stooped position

The subject was instructed to stand stooped but was allowed to bend his knees. The subject picked up the suitcase from the floor on the right side at a 20° angle using both hands and transferred it to the left in front of the body and placed it on a platform 50 cm above the floor. The platform was placed next to the subject at a 90° angle (Figure 1). This lifting technique is another option for baggage handlers inside the aircraft baggage compartment. However, this technique requires a higher ceiling in the aircraft than the kneeling position. This is why the platform height was 50 cm and not 30 cm as in the kneeling task.

Three suitcase weights of 10 kg, 15 kg and 20 kg were used and both lifting tasks were performed experimentally in a laboratory. In the analysis one trial from each task was used.

The subject practiced each task until the performance was considered consistent regarding speed and movement. The two tasks were filmed at 75 frames per second by a custom-built motion capture system of eight synchronized high speed HD cameras (GZL-CL-41C6M-C, Gazelle, Point Grey, Richmond, Canada). The subject was equipped with a full-body marker setup of 37 luminous markers with a diameter of 5 mm while three markers were placed on the suitcase.
Two force platforms (AMTI, Watertown, MA, USA) measured ground reaction forces in the standing task, while four force plates were used in the kneeling task, one under each foot and one under each knee.

6.8 Computer simulation

6.8.1 Paper II:

The models were all modifications of the "StandingModel", which is freely available in the AMMR v. 1.6.2, and were built in AMS 6.0.4. The base model was scaled to fit the bodily measures of the subject in the Wilke et al.-study (74) (72 kg, 173.9 m). Segment masses and lengths were scaled according to Winter et al (102).
The muscle redundancy problem was solved with two different criteria: 1) by minimizing the sum of muscle activities squared (2nd order polynomial) and 2) according to a minimum fatigue criterion (min/max criterion).
We compared common positions (Figure 6) in daily living (lying, sitting, standing, standing flexed) adapted from Wilke et al. (74), and since descriptions of velocities and accelerations were not provided by Wilke et al. (74), we chose to analyse the positions that were static or involved static lifting only. In the positions where the model is lying or seated, the connection between the human model and table or chair was modelled using conditional contact elements. This contact model was similar to the one published by Rasmussen et al. (103). The box had a mass of 20 kg. The output parameter (compression force) was measured in local coordinates on the cranial endplate of the L5 vertebra. The L5 endplate formed a plane to which the compression force was perpendicular.

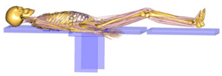
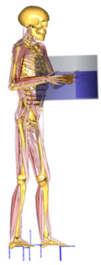
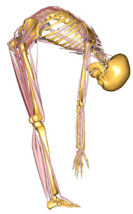
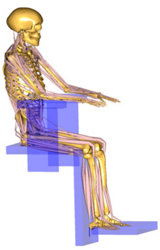
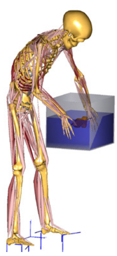
	Lying supine on a table.		Standing flexed The back in 60 degrees flexion. Arms vertical. 50 and 70 degrees of flexionare also presented.
	Sitting relaxed The back was extended 10 degrees, arm resting on armrests.		Lifting a box close to the body. The back straight and the box close to the chest.
	Standing straight The back was extended 5 degrees.		Maximum flexion The back in 95 degree flexion, hip in 50 degree flexion. Arms pointing towards toes.
	Sitting straight The back in 10 degree flexion, with no support of the back but arms resting on the armrests.		Lifting a box with stretched arms Back straight, shoulders 60 degrees flexion.
		Lifting a box with flexed back The back in 60 degrees flexion. The box 70 cm above the floor. 50 and 70 degrees of flexion are also presented.	

Figure 6. Nine different positions of the model in Paper II.

In order to compare the in vivo measurements from Wilke et al. (74) with the compression forces from the models, the spinal pressures (MPa) were converted to force (N) by:

$$F=PAC_{corr},$$

where P is the measured intra-discal pressure, A is the cross-sectional area of the L4/L5 intervertebral disc (1800 mm2) obtained from an MRI scanning and reported along with the pressure measurements (74) and $C_{corr,}$ is a correction constant of 0.77. The correction factor has shown good correlation between intra-discal pressure and compression force in a finite element model (104).

6.8.2 Paper III:

Inverse dynamics-based musculoskeletal models of the two tasks were built in the AMS v. 6.0.4. The models were modifications of the "GaitFullBody" model available from the AnyBody Managed Model Repository v. 1.5 (92) and were scaled to match the bodily measures of the subject through optimization using the method of Andersen et al.(105). The spine model consisted of seven segments (pelvis, thorax and five lumbar segments), more than 170 back and abdominal muscles parts and a model of the intra-abdominal pressure (IAP) The muscle activities were estimated according to a 2nd order polynomial optimization. This criterion proved superior in a previous validation of the lumbar spine model where it was compared with another muscle recruitment criterion (min/max) (90).

Furthermore, a suitcase-segment was added, which had the same spatial and inertial properties as the suitcase in the data collection. The model's right hand was linked to the suitcase by a revolute joint. The remaining degree of freedom was balanced by a dynamic contact model on the opposite end of the bag consisting of two contact points on the left hand and a cylindrical contact zone on the suitcase. Whenever the contact points were within the contact zone, a set of virtual muscles provided normal and frictional forces to balance the remaining degree of freedom, kinetically. This method was validated by Fluit et al. (106) for the prediction of ground reaction forces during activities of daily living. The activity of these virtual muscles was computed together with the remaining muscles in the muscle recruitment.

6.9 Statistical analysis

6.9.1 Paper I

A linear mixed model with post-hoc tukey-corrected multiple comparisons performed in SAS 9.3 (SAS institute Inc., Cary, NC, USA) was applied to identify statistically significant differences between the general and specific tasks in spinal loading and levels of muscle activity. Level of significance was set to 5 %.

6.10 Ethics

All subjects that participated in the studies involved in this thesis gave their informed consent before participation was accepted.
All parts of the study were assessed by the Regional Scientific Ethics Committee, which concluded that these studies were not notifiable (J. nr. H-3-2011-140).
The Danish Data Protection Agency allowed that data from all studies were stored (J nr. 2011-41-6915)

7 Results

7.1 Paper I

7.1.1 EMG

Relative muscular activity for all APDF levels, muscles, and tasks are presented in Table 3. In all APDF activity levels and muscles (except for the erector spinae L4/L5, P10 and trapezius, P50) the baggage compartment task had the highest level of activity. This did not reach statistical significance. In the ADPF-analysis of the full day recordings (Table 4) all activity levels were equivalent to what was found in the task-based analysis (Table 3)

Muscle	Deltoideus ant.	Deltoideus int.	Erec. Spin. L4/L5	Erec. Spin. Th12	Trapezius
P10 (%EMGmax)					
Baggage hall	0.7 (0.2)	0.6 (0.4)	3.1 (1.0)	4.1 (1.1)	2.4 (0.4)
By conveyor	0.6 (0l2)	0.8 (0.3)	4.2 (1.0)	4.5 (1.2)	1.7 (0.4)
Baggage compartment	0.6 (0.2)	0.9 (0.3)	3.5 (0.7)	6.0 (0.8)	1.5 (2.9)
P50 (%EMGmax)					
Baggage hall	3.5 (1.4)	2.8 (1.1)	8.4 (2.7)	11.8 (3.3)	7.1 (1.0)
By conveyor	3.3 (1.5)	3.5 (1.0)	12.6 (2.7)	14.5 (3.5)	6.0 (1.0)
Baggage compartment	4.5 (1.0)	4.2 (0.8)	12.9 (2.0)	18.1 (2.4)	6.6 (0.8)
P90 (%EMGmax)					
Baggage hall	19.7 (4.3)	11.8 (3.6)	21.9 (5.1)	26.8 (7.2)	17.6 (2.7)
By conveyor	18.1 (4.0)	17.3 (3.3)	33.9 (5.6)	38.7 (7.7)	20.3 (2.9)
Baggage compartment	23.2 (3.1)	19.4 (2.6)	34.9 (3.9)	41.9 (5.3)	23.6 (2.2)

Table 3. APDF in five muscles and three tasks. Mean (SE)

	Deltoideus ant.	Deltoideus int.	Erec. Spin.L4	Erec. Spin. Th12	Trapezius
P10	0.9 (0.3)	0.3 (0.04)	2.5 (0.3)	4.8 (0.9)	3.8 (2.2)
P50	6.3 (2.2)	2.6 (0.5)	9.6 (1.4)	12.3 (1.5)	11.4 (4.6)
P90	23.8 (5.5)	19.8 (3.9)	41.2 (9.3)	28.2 (2.8)	29.4 (10.5)

Table 4. APDF based on full day recordings from five muscles, but not divided into tasks. Mean (SE)

Muscle	Deltoideus ant.	Deltoideus int.	Erec. Spin.L4	Erec. Spin. Th12	Trapezius
RMS1 (%EMGmax)					
Baggage hall	99.8 (14.8)	51.1 (6.6)†	90.0 (37.1)	100.3 (32.7)	63.3 (10.0)
By conveyor	69.6 (12.6)	64.5 (6.7)	96.1 (34.3)	104.7 (34.4)	72.2 (9.4)
Baggage compartment	80.6 (12.6)	77.5 (4.7)	113.4 (24.87)	123.5 (22.6)	63.8 (6.9)
RMS5 (%EMGmax)					
Baggage hall	66.5 (10.0)	30.1 (4.0)†	50.7 (23.6)	58.1 (21.0)	40.6 (5.9)
By conveyor	41.8 (8.6)	36.7 (4.2)	56.8 (22.4)	73.8 (21.8)	44.0 (5.7)
Baggage compartment	50.2 (6.6)	48.1 (2.9)	72.1 (16.0)	79.0 (14.4)	37.9 (4.1)
RMS60 (%EMGmax)					
Baggage hall	40.3 (7.7)	13.9 (2.4)†	24.8 (11.4)	34.8 (11.5)	20.7 (3.1)
By conveyor	20.4 (6.5)	18.1 (2.4)	34.2 (10.7)	44.4 (11.5)	23.5 (3.1)
Baggage compartment	25.6 (5.1)	24.6 (1.7)	40.0 (7.7)	44.7 (7.8)	20.5 (2.2)

Table 5. Rolling RMS averages in five muscles and three tasks. †: hall ≠ compartment indicate statistically significant differences at $p < 0.05$. Mean (SE)

7.1.2 Static 2D load measurement

The L4/L5 extensor moments, compressions and shoulder moments from the general tasks are presented in Table 6 and estimates from the subtasks are presented in Table 7. The L4/L5 extensor moment in the baggage compartment task was significantly higher than in the two other tasks (Table 6). The compression force between L4 and L5 in the baggage compartment task was significantly higher than the conveyor task and the baggage hall task. There was no difference between the conveyor task and the baggage hall task (Table 6).
The biomechanical variables increased significantly ($p<0.001$) with increasing baggage weight in all tasks.
There were no significant differences in the shoulder flexor moment between the tasks.

Task/Baggage weight	10 kg	15 kg	20 kg
Compression (N/BM)			
Baggage hall	22.6 (0.5)†	27.3 (0.6)†	32.0 (0.7)†
By conveyor	21.3 (0.6)◊	26.2 (0.7)◊	31.1 (0.8)◊
Baggage compartment	29.0 (1.0)	34.1 (1.1)	39.0 (1.3)
Extensor moment (Nm/BM)			
Baggage hall	0.96 (0.03)†	1.20 (0.04)†	1.44 (0.05)†
By conveyor	0.89 (0.03)◊	1.14 (0.03)◊	1.40 (0.04)◊
Baggage compartment	1.42 (0.07)	1.70 (0.08)	1.97 (0.08)
Shoulder moment (Nm/BM)			
Baggage hall	0.24 (0.01)	0.33 (0.01)	0.43 (0.02)
By conveyor	0.26 (0.01)	0.37 (0.02)	0.48 (0.02)
Baggage compartment	0.22 (0.01)	0.33 (0.01)	0.40 (0.03)

Table 6. Compression force and extensor moment at the L4/L5 joint along with shoulder flexor moment. All are relative to body mass. †: hall ≠ compartment, ◊: conveyor ≠ compartment indicate statistically significant differences at $p < 0.05$. Mean (SE)

Task/Baggage weight	10 kg	15 kg	20 kg
Compression (N/BM)			
Loading cart	20.9 (0.82)	25.5 (0.95)	30.1 (1.1)
Unloading cart	21.9 (0.75)	27.0 (0.89)	32.0 (1.0)
Stooped	42.0 (0.96)†	47.8 (1.2)†	53.9 (1.3)†
Kneeling	26.7 (0.96)	31.8 (1.1)	36.2 (1.2)
Sitting	18.4 (1.3)	22.7 (1.6)	27.0 (1.9)
Unloading container	22.8 (1.0)	26.6 (1.3)	30.3 (1.6)
Loading container	24.9 (1.3)	30.3 (1.6)	35.7 (1.8)
Extensor moment (Nm/BM)			
Loading cart	0.87 (0.05)	1.11 (0.06)	1.35 (0.07)
Unloading cart	0.91 (0.04)	1.18 (0.05)	1.45 (0.06)
Stooped	2.40 (0.05)†	2.74 (0.07)†	3.08 (0.07)†
Kneeling	1.23 (0.06)	1.51 (0.07)	1.73 (0.08)
Sitting	0.62 (0.09)	0.87 (0.10)	1.10 (0.12)
Unloading container	1.04 (0.07)	1.24 (0.08)	1.45 (0.09)
Loading container	1.02 (0.10)	1.27 (0.13)	1.52 (0.15)
Shoulder moment (Nm/BM)			
Loading cart	0.22 (0.01)	0.32 (0.02)	0.42 (0.02)
Unloading cart	0.29 (0.02)‡	0.41 (0.02)‡	0.54 (0.03)‡
Stooped	0.12 (0.03)	0.18 (0.04)	0.24 (0.05)
Kneeling	0.26 (0.02)	0.37 (0.03)	0.45 (0.04)
Sitting	0.30 (0.03)	0.40 (0.05)	0.51 (0.06)
Unloading container	0.12 (0.02)	0.16 (0.04)	0.19 (0.04)
Loading container	0.32 (0.02)*§	0.44 (0.02)*§	0.56 (0.03)*§

Table 7. Compression force and extensor moment at the L4/L5 joint along with shoulder flexor moment for each task. All are relative to body mass. †: Stooped ≠ all other tasks, ‡: unload cart ≠ unload container, §: unloading container ≠ stooped, *: unloading container ≠ sitting indicate statistical differences at $p < 0.05$. Mean (SE)

7.2 Paper II

The measured and estimated compression forces are depicted in Figure 7. The estimated compression forces and their differences from the measured compressions are shown in Table 8.

When the 2nd order polynomial criterion for muscle recruitment was applied there was high agreement between the experimental and the modelled results. The largest absolute error was in the "sitting straight" and the "max flexed"-positions and was 176 N (resp. 29 % and 10 %) lower than in vivo data. The average relative error was 9% with the 2nd order polynomial and 16 % with the min/max criterion.). With measured values exceeding 1200 N the average error for the 2nd order polynomial was -5 % and 34 % with the min/max criterion. The largest absolute error with the min/max criterion was 831 N (33 %) in "lifting with flexed back" (Table 8).

When the compression forces were low both recruitment criteria produced comparable results, and regardless of muscle recruitment criterion the model predicted the changes in spinal compression well (Figure 7).

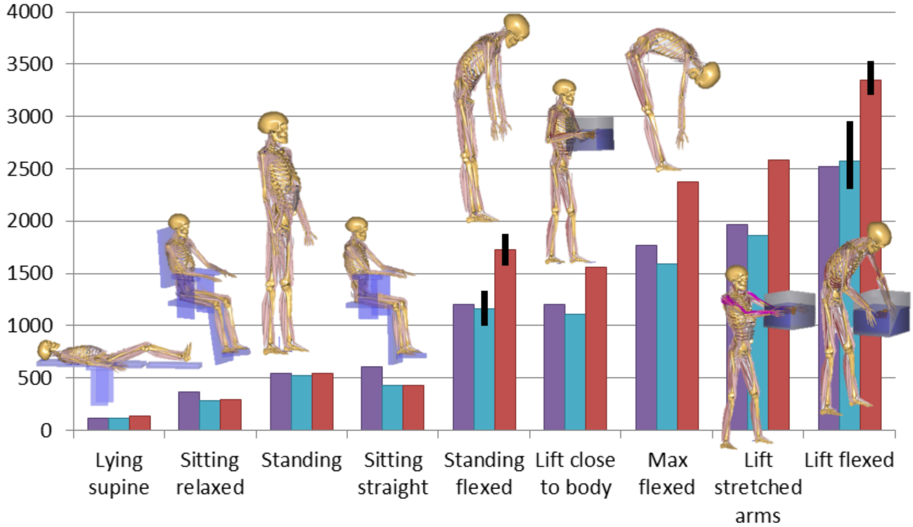


Figure 7. Estimated compression forces from the model and in vivo measurements. Purple: in vivo measurements, turquise: 2nd order polynomial, red: min/max criterion. Black bars represent compression forces in 50 and 70 degrees of flexion.

Position/ Measurement	Wilke in vivo (N)	2nd order polynomial (N)	Difference (N / %)	Min/Max-criterium (N)	Difference (N / %)
Lying supine	110	113	3 / 3	138	28 / 25
Sitting relaxed	361	281	-80 / -22	290	-71 / -20
Standing	548	518	-30 / -5	548	0 / 0
Sitting straight	602	426	-176 / -29	424	-178 / -30
Standing flexed (60°)	1205	1159	-46 / -4	1730	525 / 49
Lift close to body	1205	1104	-101 / -8	1553	348 / 29
Max flexed	1766	1590	-176 / -10	2375	609 / 34
Lift stretched arms	1971	1862	-109 / -6	2581	610 / 31
Lift flexed back (60°)	2519	2573	54 / 2	3350	831 / 33

Table 8. Absolute compression forces from two muscle recruitment criterions and the in vivo study. Error is the difference between the modeled estimate and the in vivo measurement.

Task	Weight (Kg)	Compression (N) (peak/median/ IQR)	Shear (N) (peak/ median/IQR)	Rotator moment (Nm) (peak/ median/IQR)
Kneeling	20	4197/2977/1051	237/148/71	69/9/79
Stooped	20	4692/3407/605	389/151/85	165/94/60
Kneeling	15	3341/2688/997	168/102/52	66/-2/75
Stooped	15	4801/3030/987	488/68/132	152/82/74
Kneeling	10	3039/2108/1067	125/98/70	47/-22/66
Stooped	10	5541/2740/3525	346/111/284	173/81/31

Table 9. The peak, median and inter quartile range for compression, A/P shear forces, and internal/ external rotator moment for 10 kg, 15 kg and 20 kg suitcase in the two tasks.

7.3 Paper III

The compression forces are presented in Table 9. For the 20 kg suitcase the largest compression force was found in the stooped position (4692 N) and the largest A-P shear force (289 N) also in the stooped position. For the 15 kg suitcase the largest compression force (4801 N) and A-P shear force (488 N) were also found in the stooped position. For the 10 kg suitcase the largest compression force (5541 N) and the largest A-P shear force (346 N) were found in the stooped position as well. In the stooped position, a peak of compression force occurred in the beginning of the task when the suitcase was accelerated (Figure 8). The largest peak of both compression and A-P shear forces occurred halfway through the task. This coincided with the instant at which the box was lifted off the floor. The peak compression and A-P shear forces in the kneeling position occurred in the last third of the task, where the subject lifted the suitcase towards his chest (Figure 9).

The maximal muscle force was 362 N in the right obliquus internus in the stooped position (Figure 8) and 135 N in the right obliquus externus in the kneeling position (Figure 9). In the stooped position, the first overall peak of muscle force coincided with the first peak in the compression and A-P shear force. Furthermore, the second peak of the left and right obliquus internus coincided with the largest peak of the compression force and A-P shear force (Figure 8). At the time of the overall peak of compression force the right obliquus internus also showed a peak of force.

In the kneeling position, the peak of the right obliquus internus force occurred at the same instant as the largest peak of compression force (Figure 9).

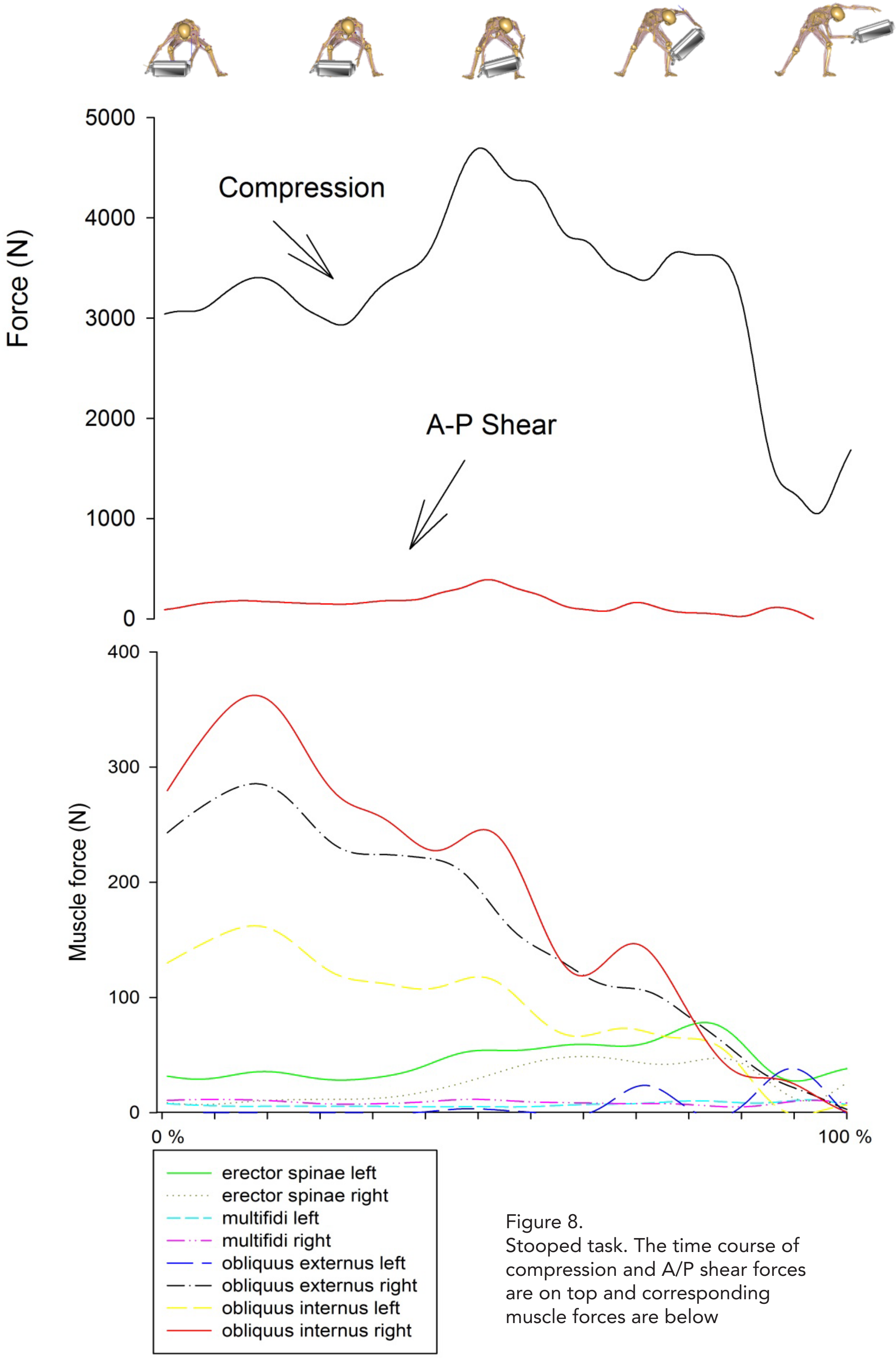


Figure 8.
Stooped task. The time course of compression and A/P shear forces are on top and corresponding muscle forces are below

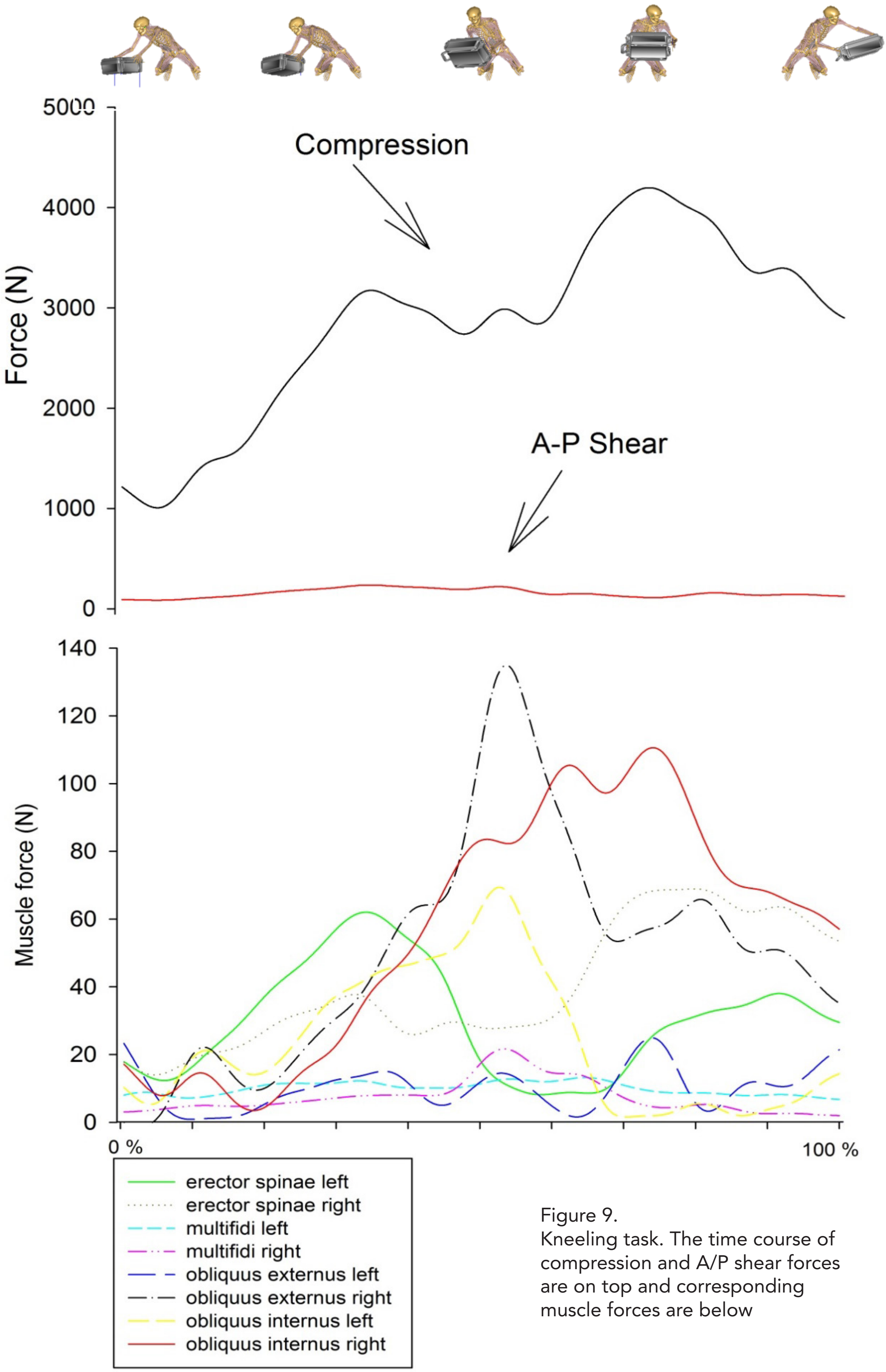


Figure 9.
Kneeling task. The time course of compression and A/P shear forces are on top and corresponding muscle forces are below

8 Discussion

This thesis aimed to describe and analyse the loading on the lumbar spine in airport baggage handlers. This was performed with a work task based approach, and the musculoskeletal loading in the different tasks will be included in the epidemiological study as exposure weights to the questionnaire and registry based data. Hence, we aimed to investigate if a dose-response relationship existed for heavy lifting and musculoskeletal pain.

The first study aimed to investigate the loading on a broad range of baggage handling tasks. This was performed with EMG measurements and static 2D load measurements. We found that the muscular activity was quite high in short periods of time, but the APDF analysis did not show remarkable levels of muscular activity. Furthermore, there were very few differences between the general work tasks in the EMG analysis. In the spinal loading estimates he level of compression force was remarkably low, in spite of high muscle activity. We found that it was significantly more loading to work in the baggage compartment than in the baggage hall and outside the aircraft by the conveyer.

The second study sought to validate the compression forces estimated with the lumbar spine model included in the AMS. This was done by comparing the compression forces in different body position with intra-discal pressures in similar position taken from the literature. We found high agreement between the model estimates and the in vivo measurements.

In the third study we used the AMS spine model to investigate two common work tasks for baggage handlers. We found that all tasks exceeded the recommended limits for compression and some approached the average maximal compression tolerance in vertebrae. Furthermore, though not in the paper, we analysed another 12 work tasks for musculoskeletal load (Appendix I).

8.1 Methodological considerations

8.1.1 Paper I

The selection of participants for the studies in Paper I was mostly random. The first 11 participants were selected by the local leader, and a date and time was agreed with the test leader. This method of recruitment led to some suspicion from the baggage handlers, who thought that the baggage handler in question would be assigned to easier tasks so the job would seem less strenuous. To counter this the authors decided that the selection of participants for the rest of the data collection should be independent of company management. We decided to show up unannounced and pick a baggage handler to test. Therefore the last 12 subjects were selected based on who would volunteer to be tested when approached on a given day.

Initially we selected 20 tasks (Table 2) that largely described the job as a baggage handler. Later we decided to collapse these 20 tasks into 3 more general tasks based primarily on where the baggage handling took place; baggage handling in the baggage hall, by the conveyer or inside the baggage compartment. The merger of these tasks could have caused us to overlook some detail, as the tasks are not necessarily comparable. If the baggage handler sits in the baggage compartment while lifting a 20 kg suitcase the compression on the L4/L5 is 27 N/BM but if the baggage handler stands stooped the compression force is 54 N/BM. And because the baggage handler does not necessarily spend equal amounts of time in each position, a simple average does not express the true loading on the lumbar spine in the general baggage compartment-task. To achieve a more valid measure of the true loading in the general task a weight for the time spend in each task could have been added. However, we are not convinced that the estimates of lumbar compression force in Paper I are valid. The calculations were performed with the Watbak-software, which provided a static 2D estimate of the L4/L5 compression force based on segment angles and the weight and direction of the burden. Because the models were two-dimensional and static, they did not take into account the movements in other than sagittal direction, nor the accelerations of the body and burden that was handled. This will most likely underestimate the compression forces and joint moments. Moreover, the model only contains one muscle producing the lumbar extensor moment with a fixed moment arm of 6 cm. This is a very crude assumption since there are many muscles balancing the extensor moment and they originate and insert at different sites, thus producing force on the lumbar spine with individually different moment arms that vary with body size. In addition, this model estimates the load on the lumbar spine on a single segment level, which does not satisfy the equilibrium at different levels of the spine. However, the method did allow us to explore differences between the tasks. Another strength of the methods in Paper I is that the measurements are from a real life setting, so it reflects a simplified version of the actual work of the baggage handlers.

8.1.2 Paper II

Generally the validation process of musculoskeletal model is very difficult. This is mostly due to the issue of retrieving valid muscle and joint forces from in vivo studies. In Paper II we compared intra-discal pressure measurements to compression forces estimated by the lumbar spine model in AMS. The conversion between force and pressure poses a potential flaw. Earlier it has been shown that a simple conversion from pressure to force (F=PA, where A is the area of the involved disc) is inadequate due to the heterogeneous material composition and therefore non-uniform loading of the disc (104;107), and will overestimate the force up to 40 % (71;104). Furthermore, during human movement the axial loading is always accompanied by shear forces and joint moments. Therefore we used a correction factor of 0.77 found in the literature (104). This correction factor is a model specific constant, and therefore probably not accurate in our case, but only in the case in which Dreischarf et al(104). introduced it. If we wanted an accurate correction factor a finite element analysis investigating the tissue-response to different types of compression in this specific model should be conducted.

The positions of the model in Paper II were all estimated based on descriptions and photographs from Wilke et al (74). The validity of the estimations would have improved markedly if kinematic data or segment/joint angles had been available. In the present case we estimated the positions, and this poses a potential bias. We showed that an estimation error of 20 degrees flexion between the pelvis and the thorax can result in estimates with an error larger than 500 N wrong (Figure 7). Also, the segment properties were estimated based on the anthropometric fractions by Winter (102), and therefore pose a potential bias, as it is uncertain if the subject in Wilke et al. (74) had a body composition that matched the general anthropometric fractions. Wilke et al. (74) did report on a variety of anthropometric parameters, but these were not applicable with the required anthropometric input in AMS.

8.1.3 Paper III and dynamic measures of musculoskeletal loading

In general, many of the issues mentioned in section 8.1.2 apply to Paper III as well. The same spinal model was applied, but the model was dynamic and driven by kinematics from the motion capture. Another limitation is the design of the study, which is based on one subject performing one trial of each baggage handling task. This limits the generalizability. However, we took measures to reduce the variation between the tasks. The subject practiced the task until the quality was considered consistent. However, this did not prove sufficient, as we have estimated larger forces in some 10 kg tasks than in the associated 20 kg tasks. This implies that the loading on the spine is not only influenced by the weight of the burden, but also indeed by the speed and accelerations of the lift.

The results from Paper III may be highly dependent of the orientation of the L5 coordinate system (Figure 10). The orientation of the coordinate system was changed to a more anatomically correct orientation. We used the current orientation, because it was validated for compression forces in Paper II (90). However, there is no report on the validity of the shear forces, joint moments or muscle forces in the present model. Therefore the sensitivity of these variables to changes in the orientation of the L5 coordinate system should be investigated in more detail. In addition, estimates of shear force, joint moments and muscle forces should be used with caution.

In the present model of the lumbar spine no ligaments are included. Instead we assumed that the joints between the vertebrae were spherical joints, hence disallowing any translations. In the human body these translations would have been limited by spinal ligaments, during which the ligaments would have contributed to the compression force. This may have caused us to underestimate the compression forces. However, the moment arm of these ligaments is very small and we assumed the contribution to be negligible.

Lastly the models are based on motion capture in a lab-setting and not a real life setting. This may further weaken the generalizability of the results, compared to a scenario where the models were based on movements recorded in the actual tasks. This could be done with the recently progressing accelerometer-based motion capture systems and the method for estimating ground reaction forces that we used in Paper II, which has previously been validated for activities of daily living (106). This would have added an extra aspect of generalizability to the results.

8.2 Discussion of findings

8.2.1 Paper I

The level of activity (APDF) in the trapezius was equivalent to the level of muscle activity in house painters in a laboratory setting (P10: 1.59 %EMGmax, P50: 6.8 % EMGmax, P90: 17.47 %EMGmax) (108). However, the painters performed intensive periods of work in different tasks as opposed to Paper I which was performed in a genuine work setting where both expected and unexpected breaks in the tasks occurred. This may also be the reason for the lack of statistical differences between the full day recordings and the task based results. We expected that the task based results would show a higher level of activity than the full day recordings, because all breaks and other types of less strenuous work tasks were included. However, the APDF analysis does not take the lengths of breaks into account. So a baggage handler performing the conveyor task could have several small periods without baggage handling, and the results from the APDF would be similar to those from a baggage handler who had a long break and then more continuous strenuous work. This means that we may have underestimated the muscle activity in the work tasks of the baggage handlers because the work task did not solely consist of the work task but involved a lot of small breaks also. However, the results do reflect the actual activity demands, as the recordings were done in the genuine work environment of the baggage handlers.

The RMS analysis showed some large muscle activity levels exceeding 100 %EMGmax. In a study of dentists Finsen & Christensen (109) found a max level of 17 %EMGmax in m. trapezius during cavity filling with a one second rolling RMS window. In comparison we found 72 %EMGmax in m. trapezius in average for baggage handler tasks. This is not surprising since the work as a baggage handler is obviously more strenuous than dentist work. However, the results from the RMS analysis did not concur with the results from the APDF. This may be due to the inability of APDF to adequately handle highly dynamic work. The APDF analysis is more suited for analysis of work with a static component, which was not the case in baggage handlers.

In the biomechanical loading analysis we found that the level of compression in the L4/L5 segment did not exceed the NIOSH recommendations of 3400 N (36) for the average baggage handler (82.6 kg) (1) in any of the general tasks. One explanation for the low level of compression force in the baggage compartment task is that this was an average of several positions including kneeling, stooped, and sitting. In the stooped task we recorded larger compression forces (4460 N), whereas the sitting task only produced around 2230 N of compression. This is not an unreasonable conclusion, as the baggage handler can switch between positions at will. In a previous study, Skotte et al (59), found compression forces of up to 4400 N during patient handling tasks, but with a dynamic 3D model. Furthermore, Granhed et al (110) found compression levels of up to 36,000 N during extremely heavy lifting with a 2D, static model. However, in a study of weightlifters the assumption of staticity and two-dimensionality is more correct that in a study of baggage handlers that perform highly dynamic and asymmetric lifts.

The low estimates of spinal loading and the high values of muscle activity in RMS1 do not correspond well. Normally high levels of muscle activity would result in high levels of compression, as the muscles compress the joints they span during contractions. The results from Paper I do not support that. However, the shortcomings of the musculoskeletal model (static, two-dimensional, single extensor muscle, single level disc equilibrium etc.) make it clear that the validity of the absolute compression estimates is not sufficient to draw any conclusion in that respect. Even though the validity of the absolute values is poor, the relative differences between the tasks can still provide knowledge. We found that the load on the lumbar spine was significantly larger in the baggage compartment task than the baggage hall and conveyer tasks. This could form the basis for recommending job rotation. However, the results from the model in Paper I are insufficient and should be supported by more valid models.

8.2.2 Paper II

In Paper II we have presented a comparison between L4/L5 intra-discal pressures measured in vivo and estimates of L4/L5 compression force from a musculoskeletal model with two different muscle recruitment criteria. When the 2nd order polynomial criterion was applied the agreement between the measured and the estimated L4/L5 compression forces was very high and errors nearly negligible (Table 8). Especially for high levels of spinal forces the relative differences between measured and estimated compression forces were small (< 10 %).

To be able to compare different positions and investigate differences in compression force, the model must be sensitive to changes in compression force between positions and tasks. In the present study, the model showed high sensitivity to the compression force between positions and a high degree of agreement with the changes in the measured intra-discal pressure. Even though the absolute errors with the min/max criterion were large, the response to changes in conditions was adequate. Even when the forces were low, the model predicted the change in the measured compression between positions fairly well.

The present validation study on the spine model shows that the 2nd order polynomial for muscle recruitment is a more appropriate recruitment criterion than the min/max criterion when the muscle forces larger than 1200 N. When the muscle forces are low the min/max and 2nd order polynomial produce the same level of compression.

In a previous comparison between the compression estimates from the AMS spinal model, Rajaee et al. (111) found good agreement with the intra-discal pressures converted to force. Rajaee et al. (111) used the min/max criterion, which we found to overestimate forces. However, there are some issues that may explain why they also found good agreement. Firstly, Rajaee et al. (111) used a different correction factor taken from Shirazi-Adl & Drouin (107), which may influence the level of estimated force markedly. Secondly, Rajaee et al. (111) also estimated positions based on photographs of the subject in Wilke et al. We have shown that an erroneous estimation of flexion angle of 20 degrees may produce errors in compression estimates larger than 500 N (Figure 2).

8.2.3 Paper III

In Paper III we described the spinal loading in two common baggage handler tasks. Enclosed in Appendix I is a supplement of force estimates for spine, shoulder, knee, and hip in 14 different baggage handler tasks with three different baggage weights (10kg, 15 kg, and 20 kg). This is, to the best of our knowledge, the first and most extensive set of modelled estimates of musculoskeletal loading in baggage handlers. Previously the kneeling (24;27;112), and stooped (57;113-115) positions have been investigated, but with models containing less detail than the one in this study. Jäger et al. (84) found in their review of the literature that the estimated in vitro average compression tolerance for lumbar segments

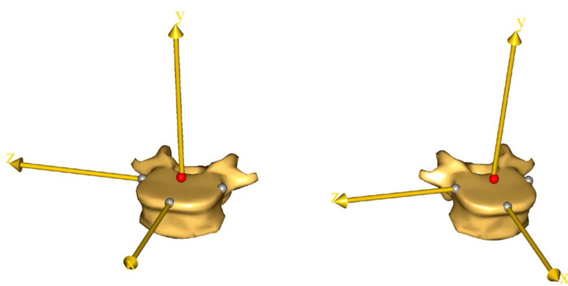


Figure 10. The anatomical reference frame. Compression force is measured in the Y-direction.

was 6180 N (SD 2660 N). Based on the compression forces from the model in the present study and the large variation of the estimate from Jäger et al. (84), compression injuries in the L4/L5 vertebrae are not unlikely to occur in baggage handling work. However, in vitro tolerance results may not be applicable for in vivo conditions. Some evidence exist that compression injuries to endplates and the underlying trabecular bone may be quite common and could be a cause for LBP (80;116). Especially large compression forces cause these injuries (60) and repeated loading increases the risk of compression injuries (60;78;82). In an in vitro study, Brinckmann et al. (78) found a 55 % risk of sustaining a compression injury if a segment was repeatedly loaded 500 times with 40-50 % of the maximum compression tolerance. This could possibly explain the high prevalence of LBP in different occupational groups with frequent heavy lifting (1;19;117). Therefore, additional mechanisms in the living organism must relieve the compression of the lumbar spine. The IAP may play an important role in reduction of the compression forces in the lumbar spine. It has been suggested that the IAP can reduce the compression force by means of a passive extensor moment (118;119). However, this will not be detectable in spinal models, which do not include a specific IAP model, as only net moments are accounted for in inverse dynamic analysis. Another possibility is that the intra-abdominal pressure acts as a semi-rigid cylinder on which the load from the upper extremities and thorax can rest (118). This will enlarge the target area for the compression from the area of the disc to the cross-sectional area of the trunk and therefore reduce the pressure on the spine markedly (118).

The shear forces found in Paper III must be considered rather modest. In a review of the literature Gallagher & Marras (86) found that appropriate limits for shear forces were 1000 N for few (<100) cycles per day and 700 N for frequent shear loading based on in vitro measurements of shear strength with no regard to other factors. Compared to these limits, a risk of developing injuries due to shear loading is not present. However, as mentioned above the results from the present study should be interpreted with caution as they may be highly dependent on the definition of the orientation of the L5 coordinate system.

9 Conclusion

In a study of muscle activity we found high levels of acute muscular activity and moderate activity over longer periods. No differences were found between the tasks regarding muscle activity. The stooped task was the most strenuous out of nine tasks measured with a static 2D model. In general the work in the baggage compartment put more load on the lumbar spine than work by the conveyer or in the baggage hall.

With a validated 3D dynamic model we elucidated the lumbar loading in two common baggage handling work tasks. We found that lifting a 10 kg suitcase in a stooped position would compress the L4/L5-joint with 5541 N. This level of compression exceeds both the NIOSH and Dortmund recommendations. Furthermore, it is close to the average vertebral compression tolerance from in vitro studies. Therefore, it is not unlikely that lifting heavy burdens in this type of positions could cause LPB.

The spine model in AMS has a unique level of detail and analysis of asymmetrical lifting tasks has not previously been carried out with a model of this level of detail. This level of detail allowed us to elucidate the lifting tasks even more realistically.

10 Perspectives

Biomechanical data with this level of detail has not previously been used as exposure measures in epidemiological studies. This method has the potential of establishing a dose-response relationship between occupational heavy lifting and musculoskeletal injuries.

The musculoskeletal models presented in Paper III are built, so they are generically usable. These models can be applied to elucidate the musculoskeletal loading in almost any lifting task. It only requires a set of kinematic data to drive the model. The ground reaction forces, that are part of the inverse dynamic analysis, can be predicted with the conditional contact model used in Paper II, so the use of and limitation by force platforms can be avoided. Therefore, data can be collected in the field during the actual work or activity. Furthermore, with this method optimization of lifting seeking to reduce spinal compression, extensor moment, muscle force etc. lifting can be performed. One drawback to this method is that the kinematics must be very accurate and with a low level of noise. Otherwise the simulated ground reaction forces will be inaccurate and the force input to the inverse dynamic analysis incorrect.

Another perspective could be to investigate the hypothesis of microfractures as a result of large spinal forces. This could be done with some of the same methods as in the present thesis. Lifting sequences recorded by a motion capture system, the kinematic data used to drive the AMS model and the ground reaction forces as input to the inverse dynamic analysis would be the initial data. Hereafter, the AMS model calculates estimates of muscle forces, joint moments and joint compressions at a certain spinal segment. These muscle and joint forces are then used as input to a Finite Element (FE) model. The FE model can be used to estimate stress level and stress distribution in the individual components (bone, discs, ligaments, etc.), and the interaction between the components (e.g. force and joint moment transfer) in the lumbar spine. The hypothesis about the development of micro fractures in the endplates of the vertebrae (60;61;120) and loosening of periost from the compact bone because of high loading can be investigated by applying specific finite element methods that can simulate the crack growth in the vertebrae. Moreover, the finite element models can also be used to simulate tissue damage. This can be done by loading the finite element model with very large muscle forces both intermittent and continuously.

11 Summary

LBP constitutes a major economic problem in many countries. The causes of LBP are still largely unknown and several risk factors have been suggested including heavy lifting, which causes high compression forces of the tissues in the low back. Micro-fractures in the endplates of the vertebrae caused by compression forces have been suggested as a source of unspecific pain. Although airport baggage handlers exhibit a high prevalence of musculoskeletal complaints the amount of biomechanical research within this and similar areas is limited. The aims of this thesis were to perform a general description of the lumbar loading in baggage handlers (Paper I), to develop a generically useful tool to examine specific lumbar compression in a valid manner (Paper II & III), and to investigate the spinal loading in common work tasks for baggage handlers. (Paper III).

We recorded electromyography during baggage handling in the baggage hall, by a conveyor, and inside the aircraft baggage compartment. Electromyography was analyzed using amplitude probability distribution functions (APDF) on both tasks and full day recordings and RMS values on tasks. Furthermore, we estimated L4/L5 compression and moment along with shoulder flexor moment with a Watbak model based on more specific subtasks. In addition, we built an inverse dynamics-based musculoskeletal computer model using the AnyBody Modeling System (AMS). Motion capture recorded the movements in 3D during a stooped and a kneeling lifting task simulating airport baggage handler work. Marker trajectories were used to drive the model. The AMS-models computed estimated compression forces, shear forces and the moments around the L4/L5 joint. The compression forces were used for comparison with the vertebral compression tolerances reported in the literature.

The RMS muscle activity was high in all tasks. The average peak RMS muscle activity was up to 120 % EMGmax in the erector spinae during the baggage hall task. There were no significant differences between the tasks in the APDF analyses. The L4/L5 compression and extensor moment from Watbak were

significantly higher in the baggage compartment task than in both the conveyor and baggage hall tasks. The stooped lifting task produced 5541 N compression in the L4/L5 joint and a kneeling task produced 4197 N in the AMS models. These compression forces were close to the average compression tolerance and exceed the recommended limits for compression during lifting.

12 Danish Summary (Dansk Resumé)

Lænderygsmerter er et stort problem I mange lande. Årsagerne til lænderyg smerter er stadig stort set ukendte og mange risikofaktorer har været nævnt, herunder tunge løft, som forårsager store kompression kræfter i vævet i lænderyggen. Mikrofrakturer i hvirvellegemernes endeplader forårsaget af store kompressions kræfter har været foreslået som en kilde til uspecifikke smerter. På trods af, at der er fundet store forekomster af muskuloskeletale smerter hos bagageportører, har mængden af biomekanisk forskning inden for området være begrænset. Formålet med denne afhandling var at lave en generel beskrivelse af belastningen på lænderyggen hos bagageportører (Paper I), at udvikle en generisk anvendeligt værktøj til at undersøge kompressions kræfter i lænderyggen validt (Paper II og III) og at undersøge belastningen på lænderyggen i udvalgte almindelige arbejdsopgaver for bagageportører (Paper III).

Vi optog elektromyografi under håndtering af bagage i bagage hallen, ved et transportbånd og i flyets lastrum. Elektromyografien blev analyseret med Amplitude Probability Distribution Functions (APDF) for både opgaver og heldagsoptagelser og med Root Mean Square (RMS) for opgaverne. Ydermere, beregnede vi kompressionen og momentet omkring L4/L5 og skulder fleksor moment med en Watbak model i mere specifikke underopgaver. Derudover byggede vi en invers dynamik baseret muskuloskeletal computer model med AnyBody Modeling System (AMS). Bevægelser i 3D blev optaget med motion capture og markørkoordinaterne blev anvendt til at drive modellen. AMS modellen beregnede estimater for kompression- og shear kræfter og ekstensor momentet i L4/L5. Kompressions kræfterne blev sammenlignet med in vitro brud grænser for hvirvellegemer fra litteraturen.

RMS muskelaktiviteten var høj i alle arbejdsopgaver. Der var ingen signifikante forskelle mellem arbejdsopgaverne i APDF-analysen. L4/L5 kompression og ekstensor moment fra Watbak var signifikant højere under arbejde i lastrummet. AMS modellen estimerede 5541 N kompression i en

foroverbøjet arbejdsopgave og 4197 N kompression under knæliggende løft. Disse kompressions kræfter var tæt på de gennemsnitlige brudgrænser for hvirvellegemer og overskred de anbefalede grænseværdier for kompression under løft.

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15 Appendix I

Overview of work tasks described in the appendix.

The Ramp
Outside the baggage compartment

1. Loading without conveyer
2. Loading with conveyer/ Unloading with conveyer/ Loading baggage-carts without lifting hook/ Unloading baggage-carts without lifting hook
3. Unloading without conveyer

Inside the baggage compartment
Loading/Unloading with conveyer in

4. Standing
5. Sitting
6. Kneeling
7. Squatting
8. Stooped

Loading/Unloading with extendible conveyer in

9. Standing
10. Kneeling
11. Squatting
12. Stooped

The Baggage hall

13. Loading baggage containers
14. Unloading baggage containers

Legend explanation:

Compression: joint compression force (N)

Abduction R/L: Shoulder abductor moment Right/Left (Nm)

Supraspinatus R/L: Supraspinatus force Right/Left (N)

Ta90 R/L: Percentage of time with shoulder above horizontal (%)

Shear: Anterior/Posterior shear (N)

Ext mom: L4/L5 extensor moment (Nm)

Rot mom: Rotator moment, counter-clock-wise positive (Nm)

Patella R/L: Patella-tendon force Right/Left (N)

Shoulder

Task	Region	Measure	peak	Median	P10	P25	P75	P90	Weight
1	shoulder	compression R	3044	1231	205	428	2754	2917	20
1	shoulder	compression L	171	84	67	74	104	129	20
1	shoulder	abduction R	41	10	2	5	20	34	20
1	shoulder	abduction L	5	2	0	1	3	3	20
1	shoulder	supraspinatus R	398	61	0	10	175	252	20
1	shoulder	supraspinatus L	14	2	0	0	5	9	20
1	shoulder	ta90 R	0						20
1	shoulder	ta90 L	0						20
1	shoulder	compression R	3471	1463	623	943	1927	2367	15
1	shoulder	compression L	230	81	45	55	117	150	15
1	shoulder	abduction R	69	18	2	7	28	35	15
1	shoulder	abduction L	4	2	1	1	2	3	15
1	shoulder	supraspinatus R	280	98	0	47	115	163	15
1	shoulder	supraspinatus L	19	0	0	0	1	8	15
1	shoulder	ta90 R	0						15
1	shoulder	ta90 L	0						15
1	shoulder	compression R	3405	1887	998	1255	2319	2659	10
1	shoulder	compression L	303	161	91	110	210	252	10
1	shoulder	abduction R	60	23	5	7	38	53	10
1	shoulder	abduction L	4	1	0	0	2	3	10
1	shoulder	supraspinatus R	405	125	66	84	185	244	10
1	shoulder	supraspinatus L	21	7	3	4	10	18	10
1	shoulder	ta90 R	0						10
1	shoulder	ta90 L	0						10
2	shoulder	compression R	2438	759	189	365	1419	1853	20
2	shoulder	compression L	313	148	82	102	210	290	20
2	shoulder	abduction R	93	34	12	26	77	87	20
2	shoulder	abduction L	4	1	0	1	2	2	20
2	shoulder	supraspinatus R	151	42	1	22	75	85	20
2	shoulder	supraspinatus L	16	6	3	4	9	14	20
2	shoulder	ta90 R	0						20
2	shoulder	ta90 L	0						20
2	shoulder	compression R	4733	1006	501	653	1387	2996	15
2	shoulder	compression L	415	208	133	158	285	353	15
2	shoulder	abduction R	75	21	10	15	32	62	15
2	shoulder	abduction L	3	1	0	0	2	3	15
2	shoulder	supraspinatus R	363	70	0	20	135	281	15
2	shoulder	supraspinatus L	19	11	6	6	15	16	15

2	shoulder	ta90 R	0					15
2	shoulder	ta90 L	0					15
2	shoulder	compression R	3452	2226	783	901	2867	3242	10
2	shoulder	compression L	427	197	124	157	340	406	10
2	shoulder	abduction R	58	35	14	22	47	52	10
2	shoulder	abduction L	3	2	1	1	3	3	10
2	shoulder	supraspinatus R	316	212	59	73	263	306	10
2	shoulder	supraspinatus L	20	14	10	12	17	19	10
2	shoulder	ta90 R	0						10
2	shoulder	ta90 L	0						10
3	shoulder	compression R	6183	3504	752	1236	3268	5103	20
3	shoulder	compression L	1492	993	390	438	1220	1299	20
3	shoulder	abduction R	2	1	0	0	2	2	20
3	shoulder	abduction L	152	92	22	53	116	138	20
3	shoulder	supraspinatus R	156	106	4	57	120	143	20
3	shoulder	supraspinatus L	175	65	6	14	85	147	20
3	shoulder	ta90 R	0						20
3	shoulder	ta90 L	0						20
3	shoulder	compression R	6176	2423	1684	2065	3159	4259	15
3	shoulder	compression L	1100	421	232	289	528	689	15
3	shoulder	abduction R	31	13	1	5	24	27	15
3	shoulder	abduction L	7	4	3	3	6	7	15
3	shoulder	supraspinatus R	1165	169	55	91	377	603	15
3	shoulder	supraspinatus L	109	31	18	20	48	77	15
3	shoulder	ta90 R	0						15
3	shoulder	ta90 L	0						15
3	shoulder	compression R	2472	430	273	305	1560	2118	10
3	shoulder	compression L	3722	1736	978	1459	2074	3256	10
3	shoulder	abduction R	17	5	2	3	10	13	10
3	shoulder	abduction L	49	11	5	6	26	41	10
3	shoulder	supraspinatus R	549	35	21	25	235	426	10
3	shoulder	supraspinatus L	382	103	67	87	152	332	10
3	shoulder	ta90 R	0						10
3	shoulder	ta90 L	0						10
4	shoulder	compression R	3348	1986	755	1137	2881	3118	20
4	shoulder	compression L	266	82	58	75	97	166	20
4	shoulder	abduction R	47	16	2	6	24	31	20
4	shoulder	abduction L	4	2	0	1	2	3	20
4	shoulder	supraspinatus R	243	104	22	69	129	159	20

4	shoulder	supraspinatus L	20	3	0	1	5	12	20
4	shoulder	ta90 R	0						20
4	shoulder	ta90 L	0						20
4	shoulder	compression R	3444	1524	905	1135	1825	2322	15
4	shoulder	compression L	620	129	69	109	211	442	15
4	shoulder	abduction R	42	6	1	3	20	28	15
4	shoulder	abduction L	7	2	0	1	2	4	15
4	shoulder	supraspinatus R	252	64	15	42	95	177	15
4	shoulder	supraspinatus L	45	4	0	0	8	29	15
4	shoulder	ta90 R	0						15
4	shoulder	ta90 L	0						15
4	shoulder	compression R	2679	1526	787	964	2060	2462	10
4	shoulder	compression L	275	142	98	123	191	215	10
4	shoulder	abduction R	44	11	1	4	17	30	10
4	shoulder	abduction L	4	1	0	1	2	3	10
4	shoulder	supraspinatus R	164	92	44	52	118	136	10
4	shoulder	supraspinatus L	26	6	3	4	10	19	10
4	shoulder	ta90 R	0						10
4	shoulder	ta90 L	0						10
5	shoulder	compression R	4713	2021	760	1435	3126	3641	20
5	shoulder	compression R	5690	1367	393	1075	2184	4368	20
5	shoulder	abduction R	46	22	15	20	33	43	20
5	shoulder	abduction L	33	12	2	9	15	16	20
5	shoulder	supraspinatus R	71	45	11	32	54	62	20
5	shoulder	supraspinatus L	41	1	0	0	21	34	20
5	shoulder	ta90 R	0						20
5	shoulder	ta90 L	0						20
5	shoulder	compression R	2736	1700	1290	1382	2374	2660	15
5	shoulder	compression L	3157	1908	1083	1238	2230	2513	15
5	shoulder	abduction R	47	20	12	14	33	44	15
5	shoulder	abduction L	39	12	9	10	16	28	15
5	shoulder	supraspinatus R	140	79	44	55	112	131	15
5	shoulder	supraspinatus L	66	0	0	0	22	43	15
5	shoulder	ta90 R	0						15
5	shoulder	ta90 L	0						15
5	shoulder	compression R	4263	2043	389	1327	2955	3730	10
5	shoulder	compression L	4556	1661	489	1139	2396	3100	10
5	shoulder	abduction R	47	18	4	11	25	39	10
5	shoulder	abduction L	62	8	3	5	34	44	10
5	shoulder	supraspinatus R	30	0	0	0	1	3	10
5	shoulder	supraspinatus L	20	0	0	0	2	13	10
5	shoulder	ta90 R	0						10
5	shoulder	ta90 L	0						10

6	shoulder	compression R	5280	1050	231	808	1195	1647	20
6	shoulder	compression L	3315	523	151	351	716	1081	20
6	shoulder	abduction R	95	11	2	5	22	37	20
6	shoulder	abduction L	22	2	0	1	5	8	20
6	shoulder	supraspinatus R	203	24	0	4	37	57	20
6	shoulder	supraspinatus L	69	0	0	0	10	32	20
6	shoulder	ta90 R	0						20
6	shoulder	ta90 L	0						20
6	shoulder	compression R	2844	1372	683	958	1632	1935	15
6	shoulder	compression L	1492	993	390	438	1220	1299	15
6	shoulder	abduction R	35	8	2	5	15	21	15
6	shoulder	abduction L	27	3	0	1	4	11	15
6	shoulder	supraspinatus R	55	0	0	0	2	7	15
6	shoulder	supraspinatus L	4	0	0	0	0	0	15
6	shoulder	ta90 R	0						15
6	shoulder	ta90 L	0						15
6	shoulder	compression R	1769	1271	725	924	1586	1695	10
6	shoulder	compression L	1504	905	337	581	1164	1440	10
6	shoulder	abduction R	69	29	2	4	57	64	10
6	shoulder	abduction L	47	15	3	5	43	45	10
6	shoulder	supraspinatus R	181	1	0	0	9	54	10
6	shoulder	supraspinatus L	82	1	0	0	11	76	10
6	shoulder	ta90 R	0						10
6	shoulder	ta90 L	0						10
7	shoulder	compression R	4686	2353	1399	1790	3049	4008	20
7	shoulder	compression L	2449	1370	557	731	1797	1931	20
7	shoulder	abduction R	95	34	8	14	61	86	20
7	shoulder	abduction L	109	16	7	10	18	75	20
7	shoulder	supraspinatus R	116	0	0	0	12	60	20
7	shoulder	supraspinatus L	76	4	0	0	18	40	20
7	shoulder	ta90 R	0						20
7	shoulder	ta90 L	0						20
7	shoulder	compression R	5950	2545	1127	1519	3450	4428	15
7	shoulder	compression L	2712	1241	496	872	1967	2237	15
7	shoulder	abduction R	36	24	2	9	31	35	15
7	shoulder	abduction L	33	17	1	4	20	28	15
7	shoulder	supraspinatus R	456	34	0	0	126	323	15
7	shoulder	supraspinatus L	273	26	0	2	44	65	15
7	shoulder	ta90 R	0						15
7	shoulder	ta90 L	0						15
7	shoulder	compression R	9006	1853	381	606	5685	7709	10
7	shoulder	compression L	2155	556	223	318	1235	1471	10
7	shoulder	abduction R	30	6	2	4	9	13	10

7	shoulder	abduction L	19	3	1	2	6	11	10
7	shoulder	supraspinatus R	561	52	0	0	315	481	10
7	shoulder	supraspinatus L	43	4	0	0	10	18	10
7	shoulder	ta90 R	0						10
7	shoulder	ta90 L	0						10
8	shoulder	compression R	8093	684	278	371	1582	2210	20
8	shoulder	compression L	4478	1339	1131	1190	2216	3747	20
8	shoulder	abduction R	81	12	0	3	35	58	20
8	shoulder	abduction L	53	6	1	3	9	13	20
8	shoulder	supraspinatus R	819	9	0	2	90	166	20
8	shoulder	supraspinatus L	389	91	51	74	121	301	20
8	shoulder	ta90 R	0						20
8	shoulder	ta90 L	37						20
8	shoulder	compression R	2234	1063	520	688	1296	1677	15
8	shoulder	compression L	3696	836	450	558	1645	2188	15
8	shoulder	abduction R	39	10	2	5	15	31	15
8	shoulder	abduction L	33	6	1	3	12	25	15
8	shoulder	supraspinatus R	156	106	4	73	123	131	15
8	shoulder	supraspinatus L	294	61	1	23	115	187	15
8	shoulder	ta90 R	0						15
8	shoulder	ta90 L	37						15
8	shoulder	compression R	2031	668	307	412	1255	1812	10
8	shoulder	compression L	2951	1641	612	1112	1996	2446	10
8	shoulder	abduction R	46	8	1	4	30	40	10
8	shoulder	abduction L	27	5	1	2	10	18	10
8	shoulder	supraspinatus R	114	5	0	1	34	57	10
8	shoulder	supraspinatus L	244	95	4	33	163	227	10
8	shoulder	ta90 R	0						10
8	shoulder	ta90 L	37						10
9	shoulder	compression R	2967	2454	1776	2063	2637	2900	20
9	shoulder	compression L	1837	1503	1208	1245	1724	1819	20
9	shoulder	abduction R	56	23	5	13	34	50	20
9	shoulder	abduction L	41	18	3	8	36	41	20
9	shoulder	supraspinatus R	114	40	0	9	81	103	20
9	shoulder	supraspinatus L	0	0	0	0	0	0	20
9	shoulder	ta90 R	0						20
9	shoulder	ta90 L	4						20
9	shoulder	compression R	1663	954	413	667	1181	1345	15
9	shoulder	compression L	1938	1242	825	1043	1519	1779	15
9	shoulder	abduction R	24	4	1	2	7	16	15
9	shoulder	abduction L	9	4	1	2	7	8	15
9	shoulder	supraspinatus R	57	9	0	2	28	49	15
9	shoulder	supraspinatus L	1	0	0	0	0	0	15

9	shoulder	ta90 R	0						15
9	shoulder	ta90 L	4						15
9	shoulder	compression R	4828	3837	2669	3233	4328	4653	10
9	shoulder	compression L	3424	2472	302	1875	2942	3133	10
9	shoulder	abduction R	45	37	12	31	41	43	10
9	shoulder	abduction L	20	10	0	2	17	19	10
9	shoulder	supraspinatus R	517	321	134	262	386	461	10
9	shoulder	supraspinatus L	306	261	23	210	282	296	10
9	shoulder	ta90 R	0						10
9	shoulder	ta90 L	4						10
10	shoulder	compression R	2900	1856	1459	1566	2228	2466	20
10	shoulder	compression L	2493	1875	1469	1572	2098	2372	20
10	shoulder	abduction R	87	16	2	6	33	53	20
10	shoulder	abduction L	54	15	3	6	26	35	20
10	shoulder	supraspinatus R	190	10	0	0	33	71	20
10	shoulder	supraspinatus L	0	0	0	0	0	0	20
10	shoulder	ta90 R	0						20
10	shoulder	ta90 L	0						20
10	shoulder	compression R	2145	893	367	857	1578	1861	15
10	shoulder	compression L	1247	750	209	489	1220	1233	15
10	shoulder	abduction R	58	15	5	7	23	42	15
10	shoulder	abduction L	45	11	2	5	18	35	15
10	shoulder	supraspinatus R	125	25	1	4	37	57	15
10	shoulder	supraspinatus L	3	0	0	0	0	1	15
10	shoulder	ta90 R	0						15
10	shoulder	ta90 L	0						15
10	shoulder	compression R	2116	1411	934	1029	1782	2042	10
10	shoulder	compression L	2413	1321	591	821	2064	2326	10
10	shoulder	abduction R	23	11	2	5	16	19	10
10	shoulder	abduction L	12	4	1	2	8	10	10
10	shoulder	supraspinatus R	128	18	1	5	68	115	10
10	shoulder	supraspinatus L	0	0	0	0	0	0	10
10	shoulder	ta90 R	0						10
11	shoulder	compression R	8718	3144	1687	2575	4717	6016	20
11	shoulder	compression L	7086	2882	2185	2463	4046	5579	20
11	shoulder	abduction R	152	66	40	47	84	119	20
11	shoulder	abduction L	117	56	39	47	85	105	20
11	shoulder	supraspinatus R	55	33	24	27	42	48	20
11	shoulder	supraspinatus L	403	111	39	57	156	301	20
11	shoulder	ta90 R	0						20
11	shoulder	ta90 L	0						20
11	shoulder	compression R	2979	1172	737	960	1774	2674	15
11	shoulder	compression L	2097	673	238	330	1508	1617	15

11	shoulder	abduction R	63	17	5	11	38	59	15
11	shoulder	abduction L	47	10	4	6	38	40	15
11	shoulder	supraspinatus R	113	5	0	0	42	52	15
11	shoulder	supraspinatus L	71	18	1	3	58	64	15
11	shoulder	ta90 R	0						15
11	shoulder	ta90 L	0						15
11	shoulder	compression R	5626	2588	1976	2167	3062	4341	10
11	shoulder	compression L	2903	1724	1251	1442	2098	2374	10
11	shoulder	abduction R	113	68	48	52	77	90	10
11	shoulder	abduction L	66	40	30	32	51	58	10
11	shoulder	supraspinatus R	152	33	15	21	50	103	10
11	shoulder	supraspinatus L	41	23	13	19	27	38	10
11	shoulder	ta90 R	0						10
11	shoulder	ta90 L	0						10
12	shoulder	compression R	5081	2516	1743	1908	3280	3987	20
12	shoulder	compression L	3004	2435	2078	2180	2803	2914	20
12	shoulder	abduction R	77	49	12	33	58	67	20
12	shoulder	abduction L	69	17	2	7	52	59	20
12	shoulder	supraspinatus R	290	8	0	0	137	254	20
12	shoulder	supraspinatus L	0	0	0	0	0	0	20
12	shoulder	ta90 R	0						20
12	shoulder	ta90 L	0						20
12	shoulder	compression R	1518	596	374	442	1114	1395	15
12	shoulder	compression L	1435	602	191	424	820	973	15
12	shoulder	abduction R	28	8	1	3	18	24	15
12	shoulder	abduction L	22	4	1	2	11	15	15
12	shoulder	supraspinatus R	119	23	0	2	69	106	15
12	shoulder	supraspinatus L	98	27	3	18	53	67	15
12	shoulder	ta90 R	0						15
12	shoulder	ta90 L	0						15
12	shoulder	compression R	3055	1907	1380	1488	2771	2955	10
12	shoulder	compression L	7336	3434	2582	2808	5549	6590	10
12	shoulder	abduction R	30	5	1	2	14	24	10
12	shoulder	abduction L	37	12	5	8	28	34	10
12	shoulder	supraspinatus R	369	196	133	149	311	347	10
12	shoulder	supraspinatus L	1395	305	208	226	737	1173	10
12	shoulder	ta90 R	0						10
12	shoulder	ta90 L	0						10
13	shoulder	compression R	7735	1873	1059	1658	2776	5774	20
13	shoulder	compression L	1825	190	63	103	264	397	20
13	shoulder	abduction R	87	20	6	12	41	70	20
13	shoulder	abduction L	11	1	0	1	2	2	20
13	shoulder	supraspinatus R	949	160	65	121	245	528	20

13	shoulder	supraspinatus L	152	12	2	3	22	37	20
13	shoulder	ta90 R	0						20
13	shoulder	ta90 L	0						20
13	shoulder	compression R	2976	1356	571	1060	1823	2308	15
13	shoulder	compression L	857	197	80	119	282	324	15
13	shoulder	abduction R	50	12	1	3	29	43	15
13	shoulder	abduction L	12	2	0	1	2	5	15
13	shoulder	supraspinatus R	268	74	5	28	155	205	15
13	shoulder	supraspinatus L	54	14	4	9	18	29	15
13	shoulder	ta90 R	0						15
13	shoulder	ta90 L	0						15
13	shoulder	compression R	4227	1340	719	944	2721	3917	10
13	shoulder	compression L	392	198	124	144	235	311	10
13	shoulder	abduction R	54	16	3	4	42	49	10
13	shoulder	abduction L	6	2	1	1	3	4	10
13	shoulder	supraspinatus R	338	96	25	40	227	310	10
13	shoulder	supraspinatus L	35	16	4	10	20	27	10
13	shoulder	ta90 R	0						10
13	shoulder	ta90 L	0						10
14	shoulder	compression R	6137	1235	712	910	1903	2475	20
14	shoulder	compression L	2602	1387	589	678	1856	2218	20
14	shoulder	abduction R	18	8	1	3	12	15	20
14	shoulder	abduction L	48	22	2	12	30	37	20
14	shoulder	supraspinatus R	1184	90	23	54	160	364	20
14	shoulder	supraspinatus L	370	184	71	81	275	323	20
14	shoulder	ta90 R	0						20
14	shoulder	ta90 L	0						20
14	shoulder	compression R	2377	1013	726	809	1272	1357	15
14	shoulder	compression L	403	260	142	188	287	353	15
14	shoulder	abduction R	26	17	2	6	19	23	15
14	shoulder	abduction L	3	2	2	2	3	3	15
14	shoulder	supraspinatus R	319	51	6	14	85	147	15
14	shoulder	supraspinatus L	23	17	8	15	19	21	15
14	shoulder	ta90 R	0						15
14	shoulder	ta90 L	0						15
14	shoulder	compression R	3284	2789	2516	2736	3018	3272	10
14	shoulder	compression L	1700	1660	1268	1509	1694	1700	10
14	shoulder	abduction R	30	27	23	25	29	29	10
14	shoulder	abduction L	25	23	19	20	24	24	10
14	shoulder	supraspinatus R	126	112	95	105	118	125	10
14	shoulder	supraspinatus L	180	169	124	151	171	178	10
14	shoulder	ta90 R	0						10
14	shoulder	ta90 L	0						10

L4/L5

Task	Region	Measure	Peak	Median	P10	P25	P75	P90	Weight
1	L4/L5	compression	3363	2252	1361	1685	2890	3116	20
1	L4/L5	shear	173	103	74	89	118	147	20
1	L4/L5	ext mom	159	83	35	54	127	146	20
1	L4/L5	rot mom	16	2	-14	-3	12	14	20
1	L4/L5	compression	2740	1908	1449	1506	2488	2665	15
1	L4/L5	shear	109	78	57	67	94	103	15
1	L4/L5	ext mom	173	98	70	83	137	160	15
1	L4/L5	rot mom	34	7	-19	-3	11	19	15
1	L4/L5	compression	3120	2058	1264	1512	2699	3016	10
1	L4/L5	shear	131	99	51	79	120	128	10
1	L4/L5	ext mom	131	38	10	19	84	116	10
1	L4/L5	rot mom	21	-5	-38	-22	5	20	10
2	L4/L5	compression	3410	2031	1355	1620	2854	3262	20
2	L4/L5	shear	157	83	41	46	136	152	20
2	L4/L5	ext mom	108	42	19	25	80	96	20
2	L4/L5	rot mom	37	8	-11	-3	23	35	20
2	L4/L5	compression	3234	2023	848	1490	2167	2587	15
2	L4/L5	shear	79	38	9	27	62	68	15
2	L4/L5	ext mom	143	53	30	33	86	122	15
2	L4/L5	rot mom	65	26	-1	10	44	56	15
2	L4/L5	compression	4243	3344	1250	1723	3676	4079	10
2	L4/L5	shear	153	87	13	29	129	146	10
2	L4/L5	ext mom	132	14	-8	0	93	126	10
2	L4/L5	rot mom	26	-23	-47	-32	23	26	10
3	L4/L5	compression	5110	3671	1379	2926	4304	4854	20
3	L4/L5	shear	597	414	216	293	509	545	20
3	L4/L5	ext mom	150	66	-63	17	113	134	20
3	L4/L5	rot mom	69	11	-15	-12	31	54	20
3	L4/L5	compression	2907	2162	1685	1763	2454	2644	15
3	L4/L5	shear	165	87	42	65	111	122	15
3	L4/L5	ext mom	167	107	70	83	136	144	15
3	L4/L5	rot mom	43	-9	-41	-26	5	20	15
3	L4/L5	compression	3966	2969	1481	2334	3239	3502	10
3	L4/L5	shear	190	112	91	98	141	170	10
3	L4/L5	ext mom	100	7	-136	-55	53	78	10
3	L4/L5	rot mom	52	30	-22	-13	49	50	10
4	L4/L5	compression	4239	2964	1859	2207	3659	4109	20
4	L4/L5	shear	193	133	72	112	176	188	20
4	L4/L5	ext mom	205	127	72	98	180	197	20
4	L4/L5	rot mom	10	-13	-24	-18	-4	4	20
4	L4/L5	compression	4869	3398	2568	3215	3625	3797	15
4	L4/L5	shear	161	107	19	61	122	143	15

4	L4/L5	ext mom	226	191	134	166	209	220	15
4	L4/L5	rot mom	166	-4	-63	-43	54	114	15
4	L4/L5	compression	3369	2261	1922	2057	2642	3290	10
4	L4/L5	shear	157	92	67	84	112	148	10
4	L4/L5	ext mom	157	99	79	89	118	154	10
4	L4/L5	rot mom	24	5	-28	-22	17	21	10
5	L4/L5	compression	4862	3597	3123	3319	4045	4297	20
5	L4/L5	shear	389	281	156	233	332	364	20
5	L4/L5	ext mom	46	3	-47	-25	32	41	20
5	L4/L5	rot mom	111	-61	-96	-82	5	71	20
5	L4/L5	compression	6641	3797	2095	2273	5338	6274	15
5	L4/L5	shear	407	190	109	135	294	369	15
5	L4/L5	ext mom	90	60	47	54	71	82	15
5	L4/L5	rot mom	154	35	-102	-10	118	139	15
5	L4/L5	compression	2819	2354	2053	2199	2448	2703	10
5	L4/L5	shear	293	210	131	174	235	279	10
5	L4/L5	ext mom	-5	-21	-35	-26	-15	-10	10
5	L4/L5	rot mom	64	-36	-41	-38	-26	23	10
6	L4/L5	compression	4197	2977	1476	2167	3140	3950	20
6	L4/L5	shear	237	148	111	125	210	226	20
6	L4/L5	ext mom	164	113	52	72	129	148	20
6	L4/L5	rot mom	69	9	-51	1	49	59	20
6	L4/L5	compression	3821	2701	1464	1623	2941	3387	15
6	L4/L5	shear	160	105	14	76	132	151	15
6	L4/L5	ext mom	175	132	-65	74	146	161	15
6	L4/L5	rot mom	66	36	-14	-2	57	129	15
6	L4/L5	compression	3317	2009	1569	1646	2087	2163	10
6	L4/L5	shear	147	101	95	99	123	144	10
6	L4/L5	ext mom	115	59	47	54	88	113	10
6	L4/L5	rot mom	47	35	-14	9	39	47	10
7	L4/L5	compression	3901	3367	3136	3284	3510	3745	20
7	L4/L5	shear	63	33	6	15	47	53	20
7	L4/L5	ext mom	171	163	142	156	166	169	20
7	L4/L5	rot mom	75	43	3	8	70	72	20
7	L4/L5	compression	5045	2825	1428	1718	3986	4609	15
7	L4/L5	shear	430	70	23	46	174	338	15
7	L4/L5	ext mom	205	162	59	65	180	198	15
7	L4/L5	rot mom	102	55	-50	-6	83	93	15
7	L4/L5	compression	3446	3088	2464	2874	3244	3367	10
7	L4/L5	shear	94	36	2	13	68	79	10
7	L4/L5	ext mom	152	131	119	124	139	148	10
7	L4/L5	rot mom	-12	-31	-44	-41	-21	-18	10

8	L4/L5	compression	4692	2997	3021	3231	4205	4495	20
8	L4/L5	shear	237	148	144	156	255	332	20
8	L4/L5	ext mom	175	139	123	125	155	161	20
8	L4/L5	rot mom	165	94	94	98	125	160	20
8	L4/L5	compression	4801	3030	2606	2889	3406	4299	15
8	L4/L5	shear	488	68	19	47	105	342	15
8	L4/L5	ext mom	190	103	174	182	190	196	15
8	L4/L5	rot mom	152	82	-12	14	90	154	15
8	L4/L5	compression	5541	2740	2258	2937	5015	5449	10
8	L4/L5	shear	346	111	19	43	277	328	10
8	L4/L5	ext mom	144	70	48	54	136	143	10
8	L4/L5	rot mom	174	81	31	76	152	167	10
9	L4/L5	compression	6288	4932	3346	4175	5674	6196	20
9	L4/L5	shear	206	176	148	156	189	200	20
9	L4/L5	ext mom	174	141	111	121	148	155	20
9	L4/L5	rot mom	84	68	39	51	79	81	20
9	L4/L5	compression	4836	3567	2721	3223	4195	4464	15
9	L4/L5	shear	235	145	108	125	183	218	15
9	L4/L5	ext mom	266	189	152	172	228	246	15
9	L4/L5	rot mom	62	25	-17	-5	45	54	15
9	L4/L5	compression	3055	2601	1396	1911	2958	3007	10
9	L4/L5	shear	136	103	67	87	109	120	10
9	L4/L5	ext mom	165	141	69	102	159	161	10
9	L4/L5	rot mom	43	23	-3	11	35	39	10
10	L4/L5	compression	4283	3383	2952	3089	3831	4109	20
10	L4/L5	shear	275	187	108	148	239	257	20
10	L4/L5	ext mom	152	81	9	24	125	142	20
10	L4/L5	rot mom	62	-3	-49	-19	51	59	20
10	L4/L5	compression	4239	2964	1859	2207	3659	3949	15
10	L4/L5	shear	193	133	72	112	176	185	15
10	L4/L5	ext mom	205	127	72	98	180	192	15
10	L4/L5	rot mom	10	-13	-24	-18	-4	3	15
10	L4/L5	compression	4650	2798	1886	2446	3983	4566	10
10	L4/L5	shear	346	116	71	78	286	331	10
10	L4/L5	ext mom	107	42	-36	-28	92	104	10
10	L4/L5	rot mom	28	-19	-30	-27	22	25	10
11	L4/L5	compression	4523	3834	3257	3633	4260	4432	20
11	L4/L5	shear	52	31	10	23	44	48	20
11	L4/L5	ext mom	208	178	133	159	190	193	20
11	L4/L5	rot mom	63	-8	-42	-33	57	61	20
11	L4/L5	compression	4523	1878	1215	1464	3596	4159	15
11	L4/L5	shear	372	54	22	31	133	267	15
11	L4/L5	ext mom	180	59	25	41	144	177	15
11	L4/L5	rot mom	133	37	-46	2	65	100	15

11	L4/L5	compression	3775	3298	2564	2900	3593	3726	10
11	L4/L5	shear	120	75	18	27	97	113	10
11	L4/L5	ext mom	142	127	97	121	136	140	10
11	L4/L5	rot mom	-16	-41	-60	-47	-33	-27	10
12	L4/L5	compression	6158	4420	2669	3640	5283	5758	20
12	L4/L5	shear	198	36	5	17	71	114	20
12	L4/L5	ext mom	199	129	50	91	172	182	20
12	L4/L5	rot mom	186	121	51	70	162	178	20
12	L4/L5	compression	3665	2824	1028	1526	2964	3413	15
12	L4/L5	shear	110	58	11	27	88	100	15
12	L4/L5	ext mom	213	180	-6	54	199	206	15
12	L4/L5	rot mom	97	36	-1	12	59	80	15
12	L4/L5	compression	4138	3835	2311	3654	3951	4049	10
12	L4/L5	shear	83	53	11	30	72	77	10
12	L4/L5	ext mom	192	59	16	30	147	183	10
12	L4/L5	rot mom	105	82	28	46	92	100	10
13	L4/L5	compression	3037	2400	1836	2087	2578	2731	20
13	L4/L5	shear	139	108	64	89	121	134	20
13	L4/L5	ext mom	128	71	46	56	99	114	20
13	L4/L5	rot mom	25	-16	-32	-23	-7	12	20
13	L4/L5	compression	3117	2057	1049	1875	2415	2568	15
13	L4/L5	shear	170	56	13	37	85	93	15
13	L4/L5	ext mom	154	90	53	69	141	152	15
13	L4/L5	rot mom	105	4	-23	-17	25	43	15
13	L4/L5	compression	2916	1764	941	1431	2276	2585	10
13	L4/L5	shear	93	63	32	43	72	81	10
13	L4/L5	ext mom	123	63	5	21	88	112	10
13	L4/L5	rot mom	17	-8	-36	-24	0	5	10
14	L4/L5	compression	2769	1959	1332	1479	2332	2641	20
14	L4/L5	shear	146	71	42	63	97	131	20
14	L4/L5	ext mom	130	73	45	59	100	125	20
14	L4/L5	rot mom	-4	-20	-53	-35	-10	-8	20
14	L4/L5	compression	2785	2140	1798	1900	2436	2604	15
14	L4/L5	shear	99	52	38	44	64	81	15
14	L4/L5	ext mom	179	130	90	114	141	155	15
14	L4/L5	rot mom	31	-8	-19	-14	3	13	15
14	L4/L5	compression	3463	3307	3086	3115	3425	3460	10
14	L4/L5	shear	70	16	6	8	36	61	10
14	L4/L5	ext mom	-134	-145	-152	-151	-136	-135	10
14	L4/L5	rot mom	-5	-13	-26	-20	-9	-6	10

Knee

Task	Region	Measure	Peak	Median	P10	P25	P75	P90	Weight
1	knee	compression R	3053	2101	520	781	2777	2984	20
1	knee	compression L	1926	579	138	214	1292	1714	20
1	knee	shear R	783	430	131	174	660	686	20
1	knee	shear L	619	247	33	67	354	498	20
1	knee	patella R	295	0	0	0	133	230	20
1	knee	patella L	727	159	10	55	348	464	20
1	knee	compression R	2288	1057	271	472	1816	2055	15
1	knee	compression L	810	617	81	157	702	720	15
1	knee	shear R	452	216	92	165	319	407	15
1	knee	shear L	239	93	11	21	163	212	15
1	knee	patella R	177	28	0	1	82	129	15
1	knee	patella L	0	0	0	0	0	0	15
1	knee	compression R	1729	615	247	323	1519	1609	10
1	knee	compression L	4246	2128	221	678	3700	3970	10
1	knee	shear R	616	209	86	108	491	524	10
1	knee	shear L	873	354	34	105	732	848	10
1	knee	patella R	622	32	0	0	379	589	10
1	knee	patella L	280	0	0	0	97	203	10
2	knee	compression R	2157	1518	410	572	1832	2132	20
2	knee	compression L	4287	1331	315	402	2970	4024	20
2	knee	shear R	533	299	117	153	423	526	20
2	knee	shear L	912	273	37	83	629	803	20
2	knee	patella R	227	10	0	0	176	204	20
2	knee	patella L	88	0	0	0	2	55	20
2	knee	compression R	1824	1413	238	310	1760	1783	15
2	knee	compression L	839	565	46	60	763	829	15
2	knee	shear R	540	492	92	123	523	535	15
2	knee	shear L	224	82	6	15	212	218	15
2	knee	patella R	615	271	0	27	479	596	15
2	knee	patella L	0	0	0	0	0	0	15
2	knee	compression R	1681	1173	214	419	1351	1641	10
2	knee	compression L	5971	1819	3	18	5558	5930	10
2	knee	shear R	437	234	74	108	299	401	10
2	knee	shear L	1001	51	4	5	896	977	10
2	knee	patella R	403	119	0	0	265	360	10
2	knee	patella L	116	1	0	0	24	29	10
3	knee	compression R	3837	2263	1479	1277	2756	3296	20
3	knee	compression L	3176	691	1501	179	1680	2428	20
3	knee	shear R	739	392	383	192	575	657	20
3	knee	shear L	1132	241	719	29	747	940	20
3	knee	patella R	183	0	11	0	11	97	20
3	knee	patella L	1647	280	948	0	948	1298	20

3	knee	compression R	4007	2489	1046	1494	3759	3915	15
3	knee	compression L	719	477	27	97	616	654	15
3	knee	shear R	933	435	192	253	794	909	15
3	knee	shear L	148	28	2	6	117	139	15
3	knee	patella R	108	0	0	0	2	42	15
3	knee	patella L	0	0	0	0	0	0	15
3	knee	compression R	2835	1877	268	1216	2318	2626	10
3	knee	compression L	2568	483	78	218	1667	2010	10
3	knee	shear R	641	384	46	187	447	585	10
3	knee	shear L	900	151	16	43	715	776	10
3	knee	patella R	161	21	0	0	81	141	10
3	knee	patella L	1456	224	0	0	675	1237	10
4	knee	compression R	3519	2881	1144	1407	3192	3454	20
4	knee	compression L	2070	725	117	160	1570	1993	20
4	knee	shear R	2603	1625	767	997	2378	2506	20
4	knee	shear L	1248	332	32	73	620	1071	20
4	knee	patella R	2295	1407	739	835	2027	2191	20
4	knee	patella L	1326	185	6	34	658	1145	20
4	knee	compression R	2881	1746	532	1114	2122	2650	15
4	knee	compression L	526	325	141	221	451	508	15
4	knee	shear R	2596	1845	373	1236	2520	2554	15
4	knee	shear L	439	259	138	183	365	394	15
4	knee	patella R	2323	1672	116	966	2216	2247	15
4	knee	patella L	0	0	0	0	0	0	15
4	knee	compression R	2698	1279	477	589	1661	2315	10
4	knee	compression L	2795	1259	332	623	1883	2672	10
4	knee	shear R	2326	914	167	186	1837	2284	10
4	knee	shear L	443	193	118	154	269	387	10
4	knee	patella R	2180	644	0	0	1737	2114	10
4	knee	patella L	120	30	0	0	92	105	10
5	knee	compression R	1463	203	32	50	509	1126	20
5	knee	compression L	1554	1250	270	321	1421	1457	20
5	knee	shear R	175	21	3	7	100	144	20
5	knee	shear L	175	93	27	49	123	141	20
5	knee	patella R	248	50	7	27	171	208	20
5	knee	patella L	116	0	0	0	55	89	20
5	knee	compression R	360	143	88	115	198	283	15
5	knee	compression L	26	17	5	9	21	25	15
5	knee	shear R	70	13	2	6	18	49	15
5	knee	shear L	39	21	8	14	30	36	15
5	knee	patella R	79	2	0	0	19	51	15
5	knee	patella L	0	0	0	0	0	0	15
5	knee	compression R	294	102	29	44	173	186	10
5	knee	compression L	792	569	316	382	644	737	10

5	knee	shear R	59	7	2	4	13	21	10
5	knee	shear L	71	20	6	12	38	60	10
5	knee	patella R	97	20	1	10	27	33	10
5	knee	patella L	0	0	0	0	0	0	10
6	knee	compression R	2730	1400	163	261	2120	2599	20
6	knee	compression L	1738	221	21	119	346	1167	20
6	knee	shear R	318	1	-104	-94	164	271	20
6	knee	shear L	-17	-109	-146	-132	-84	-65	20
6	knee	patella R	-116	-381	-609	-472	-281	-236	20
6	knee	patella L	1	-1	-714	-104	0	0	20
6	knee	compression R	2635	912	66	121	1453	2058	15
6	knee	compression L	1247	95	16	30	203	801	15
6	knee	shear R	29	-61	-103	-80	-44	-25	15
6	knee	shear L	67	-171	-315	-235	-111	-73	15
6	knee	patella R	-94	-658	-865	-811	-213	-109	15
6	knee	patella L	1	-211	-491	-451	-4	-1	15
6	knee	compression R	2703	1335	835	1155	2375	2669	10
6	knee	compression L	297	97	26	31	167	290	10
6	knee	shear R	103	24	-129	-47	62	93	10
6	knee	shear L	-98	-134	-180	-175	-114	-101	10
6	knee	patella R	-467	-578	-720	-651	-492	-473	10
6	knee	patella L	7	0	-119	-54	0	1	10
7	knee	compression R	1749	439	120	331	807	1291	20
7	knee	compression L	1424	349	34	108	1042	1310	20
7	knee	shear R	1240	710	511	622	1001	1214	20
7	knee	shear L	2850	1316	796	864	2311	2751	20
7	knee	patella R	2712	973	696	755	2017	2652	20
7	knee	patella L	2788	1673	836	895	2165	2581	20
7	knee	compression R	661	430	156	241	532	597	15
7	knee	compression L	168	130	68	96	149	160	15
7	knee	shear R	572	387	182	238	464	501	15
7	knee	shear L	677	592	453	536	621	645	15
7	knee	patella R	1141	561	249	400	743	950	15
7	knee	patella L	0	0	0	0	0	0	15
7	knee	compression R	709	568	318	422	610	627	10
7	knee	compression L	577	130	10	64	256	331	10
7	knee	shear R	460	307	165	208	324	354	10
7	knee	shear L	1590	1415	1340	1370	1450	1506	10
7	knee	patella R	867	561	317	430	580	691	10
7	knee	patella L	2234	1667	1489	1507	1941	2060	10
8	knee	compression R	674	354	242	278	493	663	20
8	knee	compression L	3549	2876	2445	2532	3231	3457	20
8	knee	shear R	1609	563	36	170	931	1489	20
8	knee	shear L	2119	1909	1545	1626	2032	2104	20

8	knee	patella R	1388	499	19	170	794	1282	20
8	knee	patella L	1654	1476	993	1212	1607	1635	20
8	knee	compression R	2098	398	156	194	1281	1752	15
8	knee	compression L	3389	1972	1122	1339	2989	3082	15
8	knee	shear R	2234	936	265	410	1640	1974	15
8	knee	shear L	1354	1120	1003	1052	1234	1313	15
8	knee	patella R	1880	786	216	357	1367	1656	15
8	knee	patella L	1010	868	702	784	945	973	15
8	knee	compression R	1388	137	79	93	402	927	10
8	knee	compression L	3766	2787	1674	2224	3473	3709	10
8	knee	shear R	1502	162	68	81	624	985	10
8	knee	shear L	1498	1094	911	989	1417	1466	10
8	knee	patella R	1236	116	40	75	503	801	10
8	knee	patella L	1341	755	528	626	1256	1310	10
9	knee	compression R	4095	1375	343	391	2978	3758	20
9	knee	compression L	3988	2631	1124	1820	3119	3754	20
9	knee	shear R	2520	1095	71	98	2257	2479	20
9	knee	shear L	2203	1209	442	695	1830	2101	20
9	knee	patella R	1967	792	0	0	1766	1936	20
9	knee	patella L	2	0	0	0	0	0	20
9	knee	compression R	3953	3345	1436	2377	3630	3933	15
9	knee	compression L	787	39	5	13	124	446	15
9	knee	shear R	2591	2167	1851	2025	2291	2527	15
9	knee	shear L	236	23	8	17	33	126	15
9	knee	patella R	2173	1938	1755	1860	2046	2123	15
9	knee	patella L	0	0	0	0	0	0	15
9	knee	compression R	3506	2909	1585	2052	3300	3490	10
9	knee	compression L	591	397	257	313	481	532	10
9	knee	shear R	1987	1605	1120	1310	1794	1979	10
9	knee	shear L	160	128	48	85	140	150	10
9	knee	patella R	1508	1350	1187	1284	1378	1437	10
9	knee	patella L	139	0	0	0	107	132	10
10	knee	compression R	2201	94	41	63	661	1556	20
10	knee	compression L	2189	192	29	38	1625	2078	20
10	knee	shear R	68	30	17	22	43	60	20
10	knee	shear L	507	221	131	150	359	463	20
10	knee	patella R	766	154	82	96	444	689	20
10	knee	patella L	768	440	4	240	634	736	20
10	knee	compression R	674	354	242	278	493	663	15
10	knee	compression L	3549	2876	2445	2532	3231	3457	15
10	knee	shear R	1609	563	36	170	931	1489	15
10	knee	shear L	2119	1909	1545	1626	2032	2104	15
10	knee	patella R	742	373	109	231	493	549	15
10	knee	patella L	1274	198	10	61	680	1100	15

10	knee	compression R	612	209	75	111	297	413	10
10	knee	compression L	1205	708	230	480	971	1174	10
10	knee	shear R	386	120	38	52	198	279	10
10	knee	shear L	1096	340	32	80	812	1078	10
10	knee	patella R	810	297	125	158	481	669	10
10	knee	patella L	1405	694	482	528	1095	1380	10
11	knee	compression R	762	179	45	75	317	640	20
11	knee	compression L	1544	623	272	532	927	1209	20
11	knee	shear R	964	790	204	514	914	949	20
11	knee	shear L	2304	1767	625	850	2153	2239	20
11	knee	patella R	1951	893	112	502	1575	1700	20
11	knee	patella L	2361	1863	756	1004	2057	2226	20
11	knee	compression R	1076	426	127	169	667	949	15
11	knee	compression L	147	71	5	26	113	142	15
11	knee	shear R	905	337	109	172	584	785	15
11	knee	shear L	765	676	358	515	742	757	15
11	knee	patella R	1814	573	33	114	1240	1657	15
11	knee	patella L	0	0	0	0	0	0	15
11	knee	compression R	1149	468	309	415	615	1025	10
11	knee	compression L	1065	244	61	116	594	947	10
11	knee	shear R	655	342	243	276	393	566	10
11	knee	shear L	1515	1252	1188	1202	1330	1497	10
11	knee	patella R	1619	517	347	408	652	1329	10
11	knee	patella L	2225	1699	1518	1633	2062	2158	10
12	knee	compression R	2800	324	115	147	1209	2114	20
12	knee	compression L	4080	2613	1237	1333	3672	3922	20
12	knee	shear R	1920	175	32	93	1026	1678	20
12	knee	shear L	1739	1077	741	917	1348	1678	20
12	knee	patella R	1637	131	7	59	856	1411	20
12	knee	patella L	1574	649	0	5	1045	1525	20
12	knee	compression R	1554	457	111	166	809	1073	15
12	knee	compression L	489	452	361	404	475	483	15
12	knee	shear R	2919	495	67	115	1640	2344	15
12	knee	shear L	664	585	329	484	631	651	15
12	knee	patella R	2523	414	2	28	1394	2014	15
12	knee	patella L	0	0	0	0	0	0	15
12	knee	compression R	2320	214	140	165	933	1689	10
12	knee	compression L	3830	1678	913	1261	2851	3805	10
12	knee	shear R	2560	677	175	219	1817	2187	10
12	knee	shear L	1948	1379	1175	1235	1504	1850	10
12	knee	patella R	2193	578	148	189	1555	1874	10
12	knee	patella L	1582	1007	598	805	1086	1471	10
13	knee	compression R	2239	1767	659	1114	1907	2162	20
13	knee	compression L	2528	624	200	320	1987	2371	20

13	knee	shear R	1095	579	328	388	816	912	20
13	knee	shear L	1697	186	15	60	1053	1483	20
13	knee	patella R	742	373	109	231	493	549	20
13	knee	patella L	1274	198	10	61	680	1100	20
13	knee	compression R	2439	1342	473	760	1588	2113	15
13	knee	compression L	788	414	73	222	613	673	15
13	knee	shear R	994	533	281	396	785	877	15
13	knee	shear L	295	63	5	16	116	169	15
13	knee	patella R	700	394	148	231	567	618	15
13	knee	patella L	0	0	0	0	0	0	15
13	knee	compression R	1804	1031	312	530	1333	1580	10
13	knee	compression L	3414	1037	393	758	2226	2397	10
13	knee	shear R	678	486	190	299	641	661	10
13	knee	shear L	650	232	65	139	503	586	10
13	knee	patella R	769	461	120	222	601	686	10
13	knee	patella L	568	145	0	0	323	492	10
14	knee	compression R	825	371	135	301	641	756	20
14	knee	compression L	3418	2486	2085	2261	2922	3212	20
14	knee	shear R	394	219	36	139	309	361	20
14	knee	shear L	1955	1120	766	807	1875	1908	20
14	knee	patella R	295	171	2	47	207	248	20
14	knee	patella L	2101	799	430	517	1959	1991	20
14	knee	compression R	2667	1767	1235	1444	1995	2469	15
14	knee	compression L	661	327	75	99	489	618	15
14	knee	shear R	1363	1029	658	913	1219	1310	15
14	knee	shear L	143	87	2	17	123	135	15
14	knee	patella R	1709	1273	661	991	1413	1648	15
14	knee	patella L	0	0	0	0	0	0	15
14	knee	compression R	1586	1290	1213	1231	1508	1584	10
14	knee	compression L	1150	999	674	813	1074	1132	10
14	knee	shear R	655	464	366	377	611	652	10
14	knee	shear L	438	425	304	356	433	436	10
14	knee	patella R	973	697	421	495	917	973	10
14	knee	patella L	756	741	546	645	749	754	10

Hip

Task	Region	Measure	Peak	Median	P10	P25	P75	P90	Weight
1	hip	compression R	3423	2173	531	714	2534	3173	20
1	hip	compression L	1759	736	442	555	1074	1636	20
1	hip	compression R	2917	1181	244	378	2037	2674	15
1	hip	compression L	729	607	73	174	631	691	15
1	hip	compression R	1804	890	358	425	1427	1736	10
1	hip	compression L	3002	1682	467	1011	2567	2879	10
2	hip	compression R	2288	1462	736	916	1777	2173	20
2	hip	compression L	3219	1065	356	550	2617	3050	20
2	hip	compression R	2113	1011	105	291	1622	2047	15
2	hip	compression L	962	664	19	234	791	915	15
2	hip	compression R	2244	1205	731	938	1582	1897	10
2	hip	compression L	4317	2462	269	413	3983	4243	10
3	hip	compression R	3799	2416	1161	1687	2847	3323	20
3	hip	compression L	4766	1290	2296	757	3053	3909	20
3	hip	compression R	3422	2652	1178	1957	3052	3288	15
3	hip	compression L	642	404	32	83	536	580	15
3	hip	compression R	3460	2256	496	2001	2508	2832	10
3	hip	compression L	2279	1388	448	670	1767	1981	10
4	hip	compression R	4246	3407	908	1157	3719	4141	20
4	hip	compression L	1875	839	629	684	1664	1783	20
4	hip	compression R	3665	2582	868	1561	3286	3509	15
4	hip	compression L	545	387	99	204	455	487	15
4	hip	compression R	3027	1543	472	528	1778	2629	10
4	hip	compression L	2699	1875	703	1476	2167	2563	10
5	hip	compression R	3396	986	558	750	1592	2637	20
5	hip	compression L	3031	2241	1964	2111	2558	2775	20
5	hip	compression R	1794	737	484	556	937	1396	15
5	hip	compression L	901	457	131	186	696	792	15
5	hip	compression R	1536	692	563	602	884	1075	10
5	hip	compression L	1837	1589	1131	1295	1739	1807	10
6	hip	compression R	4943	3401	586	1649	4384	4845	20
6	hip	compression L	3895	832	435	712	1972	3622	20
6	hip	compression R	5420	2806	368	448	3865	4722	15
6	hip	compression L	3561	1854	610	983	3085	3487	15
6	hip	compression R	4973	2861	2525	2705	4266	4916	10
6	hip	compression L	1372	889	564	676	927	1135	10
7	hip	compression R	3673	1185	646	851	2997	3487	20
7	hip	compression L	3738	2834	1959	2083	3479	3573	20
7	hip	compression R	2399	1469	610	903	1541	1949	15
7	hip	compression L	326	63	23	41	107	200	15
7	hip	compression R	1105	792	442	585	1044	1095	10
7	hip	compression L	3864	3161	2938	3025	3474	3787	10

8	hip	compression R	1824	810	537	661	1158	1780	20
8	hip	compression L	7077	6242	4296	5138	6756	7024	20
8	hip	compression R	3391	689	210	291	2249	2921	15
8	hip	compression L	6477	4517	3538	3876	5426	5988	15
8	hip	compression R	2503	428	158	251	851	1761	10
8	hip	compression L	7611	5568	4196	4348	7084	7480	10
9	hip	compression R	4586	2133	871	972	3774	4340	20
9	hip	compression L	5503	3383	1094	1871	4510	5275	20
9	hip	compression R	4559	3806	2508	3401	4243	4457	15
9	hip	compression L	998	260	94	173	317	563	15
9	hip	compression R	3912	3315	1478	2291	3720	3835	10
9	hip	compression L	720	419	322	339	536	605	10
10	hip	compression R	5143	700	368	410	2194	3852	20
10	hip	compression L	3980	3227	921	2140	3395	3938	20
10	hip	compression R	1372	889	564	676	927	1135	15
10	hip	compression L	1536	692	563	602	884	1075	15
10	hip	compression R	852	425	362	380	551	815	10
10	hip	compression L	3338	2452	2279	2365	2976	3243	10
11	hip	compression R	3379	1947	580	653	3063	3231	20
11	hip	compression L	4843	3032	1134	1775	4464	4771	20
11	hip	compression R	3903	811	144	310	1982	2880	15
11	hip	compression L	168	42	10	22	69	129	15
11	hip	compression R	1264	921	573	683	1039	1178	10
11	hip	compression L	3444	3044	2588	2811	3230	3370	10
12	hip	compression R	3863	726	437	502	2078	3232	20
12	hip	compression L	8404	5438	3979	4878	6680	7946	20
12	hip	compression R	3037	647	260	361	1456	2209	15
12	hip	compression L	476	437	315	389	451	464	15
12	hip	compression R	3680	804	387	523	1830	2926	10
12	hip	compression L	6478	5114	2818	4870	5909	6414	10
13	hip	compression R	2885	1689	594	983	1936	2469	20
13	hip	compression L	2464	1406	773	911	1999	2333	20
13	hip	compression R	2241	1451	819	999	1529	1803	15
13	hip	compression L	1332	453	33	217	566	740	15
13	hip	compression R	2457	1285	564	949	1434	1720	10
13	hip	compression L	2508	1147	713	919	2218	2383	10
14	hip	compression R	839	488	246	338	608	661	20
14	hip	compression L	3994	2973	2163	2411	3394	3757	20
14	hip	compression R	4171	2254	1267	1911	3135	3973	15
14	hip	compression L	592	249	5	26	415	547	15
14	hip	compression R	2740	2415	2123	2165	2684	2726	10
14	hip	compression L	1757	1480	1121	1326	1590	1726	10

16 Papers

Paper I
Local muscle load and low back compression forces evaluated by EMG and video recordings of airport baggage handlers.
Henrik Koblauch, Simon Falkerslev, Stine Hvid Bern, Tine Alkjær, Charlotte Brauer, Sigurd Mikkelsen, Mark de Zee, Lau C. Thygesen, and Erik B. Simonsen.

(Draft)

Paper II
The validation of a musculoskeletal model of the lumbar spine.
Henrik Koblauch, Michael Skipper Andersen, Mark de Zee, John Rasmussen, Tine Alkjær, Charlotte Brauer, Sigurd Mikkelsen, Lau C. Thygesen, Sylvain Carbes, and Erik B. Simonsen,

(Submitted to Journal of Biomechanics)

Paper III
Spinal loads in asymmetrical and dynamic lifting tasks: A modeling approach.
Henrik Koblauch, Michael Skipper Andersen, Tine Alkjær, Charlotte Brauer, Sigurd Mikkelsen, Mark de Zee, Lau C. Thygesen, and Erik B. Simonsen

(Submitted to Journal of Applied Ergonomics)

Paper I

Local muscle load and low back compression forces evaluated by EMG and video recordings of airport baggage handlers.

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Abstract

Although airport baggage handlers exhibit a high prevalence of musculoskeletal complaints the amount of biomechanical research within this and similar areas is limited. Therefore, the purpose of this study was to quantify the musculoskeletal loading. In addition, the purpose was to investigate if significant differences existed between three work tasks.

We recorded electromyography during baggage handling in the baggage hall, by a conveyor, and inside the aircraft baggage compartment. Electromyography was analyzed using amplitude probability distribution functions (APDF) on both tasks and full day recordings and RMS values on tasks. Furthermore, we estimated L4/L5 compression and moment along with shoulder flexor moment. The only differences between the tasks in muscle activity were in the intermediate deltoid in the RMS analysis. However, the muscle activity was very high in all tasks. The average peak RMS muscle activity was up to 120 % EMGmax in the erector spinae during the baggage hall task. There were no significant differences in the APDF analyses. The L4/L5 compression and extensor moment were significantly higher in the baggage compartment task than in both the conveyor and baggage hall tasks. There was no difference in shoulder flexor moment.

In conclusion, the levels of muscular activity exceeded 100 %EMGmax in the RMS analysis. There were very few statistically significant differences between the tasks in the muscular activity. However, the baggage compartment task put significantly more load on the lumbar spine than the other tasks. In the future this work can be used to influence job rotations.

Keywords

Electromyography, compression forces, musculoskeletal loading, occupational health, heavy lifting

Abbreviations

Electromyography	EMG
Maximal EMG amplitude	EMGmax
Amplitude Probability Distribution Function	APDF
Root Mean Square	RMS

Introduction

Musculoskeletal pain constitutes a major occupational health hazard in highly developed industrialised societies (Christensen 2012; Maniadakis 2000). This is not only a problem for the single worker; it is also a massive economic issue. Measured in Disability Adjusted Life Years low back pain is a larger health burden than chronic obstructive pulmonary disease and ischemic heart disease in Western Europe (WHO, http://ihmeuw.org/25lz). In a general working population a prevalence of low back pain around 25 % has been reported by (Andersen 2007). This is less than the level found in a cohort study of musculoskeletal complaints in baggage handlers in Copenhagen Airport, where Bern et al. (2013) reported 33 % with complaints of low back pain in the past year, which was significantly higher than in a comparable control population (23 %).

Frequent and heavy lifting has been suggested as predictors for development of low back pain (Andersen 2007; Harkness 2003; Latza 2000). This is probably caused by the high mechanical loading on the lumbar spine caused by lifting (Koblauch 2013). Furthermore, work in non-neutral positions and in constrained spaces has been related to increased loading of the low back (Gallagher 2005; Stalhammar 1986) and asymmetrical lifting has been shown to increase spinal loading (Gallagher 1994). An important factor that contributing to mechanical loading of the skeletal system is muscle activity. High levels of muscle activity increases the bone-on-bone forces in the joints spanned by the muscles. A continuously high level of muscle activity is considered a risk of musculoskeletal complaints (Silverstein 1986; Veiersted 1993). The airport baggage handler lifts an average of 5 tons per day with a mean baggage weight of 15 kg (Brauer 2013) and often performs lifts in awkward positions inside the aircraft baggage compartments (Oxley 2009; Tapley 2005). Although airport baggage handlers exhibit a high prevalence of musculoskeletal complaints (Kuorinka 1987; Stalhammar 1986; Undeutsch 1982a; Undeutsch 1982b) the amount of biomechanical research within this and similar areas is limited. Only a few studies have investigated the biomechanical loading to which airport baggage handlers are subjected (Jørgensen 1987; Ruckert 1992; Splittstoesser 2007; Stalhammar 1986). Detailed knowledge of the mechanical loads is

imperative to improve prevention of musculoskeletal injuries.
The aim of the present study was to further investigate musculoskeletal load during heavy lifting in airport baggage handlers in their genuine work environment with special focus on elucidating the differences between work tasks. The hypothesis was that the musculoskeletal load is higher during work in the baggage compartment than in the baggage hall or outside the aircraft. This was assumed because of the limited space in the baggage compartment that forces the baggage handlers to use awkward work positions during baggage handling.

Materials & Methods

Work task description

First, we observed baggage handlers working in the airport during a two week period and interviewed twelve of the baggage handlers about their work. Based on this information baggage handler work tasks were divided into work in the baggage hall and work on the ramp. Work in the baggage hall consisted of loading and unloading of baggage containers and belly-carts with baggage to or from a belt conveyer. A pneumatic lifting hook was available for belly-cart and open-roofed container work but could not be used with fixed-roofed containers. Work on the ramp consisted of work on the ground and work inside the airplane baggage compartments.

On the ground the work was loading and unloading belly-carts with baggage to or from a belt conveyer that transported baggage between the airplane baggage compartment opening and the belly-cart on the ground. If the aircraft baggage compartment opening was low the baggage was lifted directly to or from the opening without using a conveyer. Inside the baggage compartment the work consisted of lifting the baggage to or from the ground-to-airplane conveyer and to pack or unpack the baggage inside the compartment. Some belt conveyers were extendible and flexible allowing the baggage to be conveyed to any place in the compartment (RampSnake®, Power Stow®). Depending on the size of the compartment and conveyer belt system, loading and unloading work inside the compartment was done by one or two baggage handlers. Work positions depended on the height of the compartment

The Ramp

Outside the baggage compartment

1. Loading without conveyer
2. Loading with conveyer
3. Unloading without conveyer
4. Unloading with conveyer

Inside the baggage compartment

Loading/Unloading with conveyer in

1. Standing
2. Sitting
3. Kneeling
4. Squatting
5. Stooped

Loading/Unloading with extendible conveyer in

10. Standing
11. Sitting
12. Kneeling
13. Squatting
14. Stooped

The Baggage hall

15. Loading baggage containers
16. Unloading baggage containers
17. Loading baggage-carts without lifting hook
18. Loading baggage-carts and open-roof containers with lifting hook
19. Unloading baggage-carts without lifting hook
20. Unloading baggage-carts and open-roof containers with lifting hook

Table 2.
Overview of the 20 general work tasks for baggage handlers.

relative to the height of the baggage handler and personal preferences, and were divided into standing, stooped, sitting, squatting and kneeling positions. From these basic work characteristics we defined 20 specific work tasks (Table 1). It was decided to collapse the 20 work tasks into 3 more general tasks: "The baggage hall", "By the conveyor", and "Inside the baggage compartment". The reduction was based on work tasks being very similar, being unmeasurable and a general question of resources. Loading and unloading at the conveyer outside the aircraft and in the baggage hall were considered to be similar. Loading and unloading with a pneumatic lifting hook were considered unmeasurable in the static computer model, as the load is carried by the hook. However, it was still a part of the baggage hall task in the EMG study, but was performed rarely, as most baggage handlers did not use the lifting hook regularly. The loading and unloading without conveyer outside the aircraft were excluded because the tasks were relatively rare, and we did not succeed in collecting sufficient data from these tasks.

After this reduction the "baggage hall" task consisted of loading and unloading belly-carts and containers, the "conveyor" task consisted of loading and unloading belly carts, and the "baggage compartment" task consisted of baggage handling in sitting, kneeling and stooped positions inside the baggage compartment (Figure 1). We did not distinguish between use of extendible conveyer in any task. For overview reasons, we report on the forces from all subtasks. The baggage handlers were specifically asked not to take notice to the measurements and perform their job as they would usually do.

Figure 1. Examples of work task performed by baggage handlers
in Copenhagen Airport. Top left: Baggage hall task, Top right:
The conveyer-task, Bottom left: Kneeling, Bottom right: stooped
positions in the baggage compartment task.

Subjects

A total of 23 baggage handlers, 39.6 years of age (range 24-56), were recruited for the EMG study. The first 11 subjects were selected by the nearest department leader. The remaining 12 were approached directly at the beginning of the workday and if the baggage handler agreed to participate he was included in the study. Full day EMG-measurements were obtained from the first 11 participants. In average the full day measurements lasted 4.6 (SD 1.2) hours. This was due to loss of data, mounting of equipment, termination of the workday due to injury and short shifts. The 11 full day measurements were from four baggage handlers on international ramp, two on domestic ramp, and two from the baggage hall. The task specific measurements were from seven baggage handlers on the international ramp and five from the baggage hall. There were no task specific measurements from the domestic ramp. In total the 23 participants contributed with a total of 102 task specific measurements, divided on 47 from baggage compartment, 19 measurements from the conveyer task and 36 from the baggage hall. In average the baggage hall tasks lasted (mean(SD)) 28.2 (14.0) minutes, the conveyer task 19.3 (13.0) minutes, and the baggage compartment task 22.6 (17.5) minutes. In the study of 2D static loading 10 baggage handlers were filmed in each sub task, and some were filmed in several tasks, so a total of 44 baggage handlers (40.2 years, 82.6 kg, 180.0 cm) participated. The authors recruited baggage handlers directly while they were performing the desired task. This method was mainly based on chance, and whoever performed a desired task was approached and asked to participate in the study.

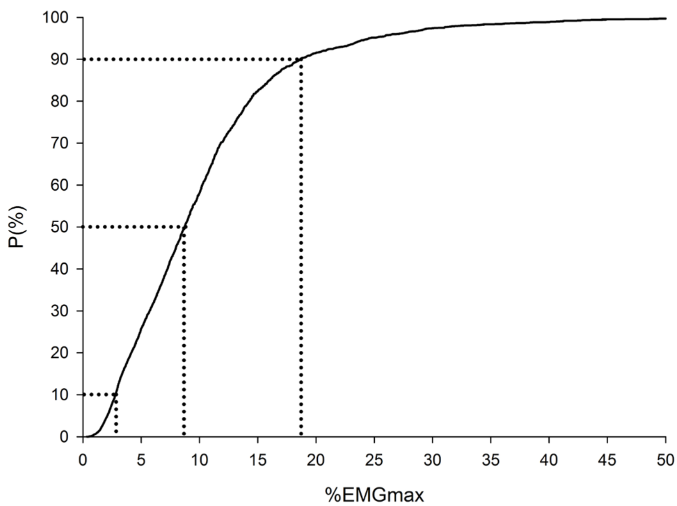

Figure 2. Example of an APDF-curve obtained from m. deltoideus intermedius. Lines show the levels p10, p50 and p90.

Nine baggage handlers participated in both parts of the study, but this did not influence the performance in either studies. The study was approved by the local ethics committee (J. nr. H-3-2011-140). Informed consent was obtained from all individual participants included in the study.

Electromyography

Bipolar EMG-electrodes (Multi Bio Sensors, Texas, USA) with a fixed interelectrode distance of 20 mm were placed on five sites on the right side: 1) m. deltoideus anterior part, 2) m. deltoideus intermediate part, 3) m. erector spinae at L4/L5-level, 4) m. erector spinae at Th12-level, and 5) descending part of m. trapezius. A reference electrode was placed on the processus spinosus of C7. Prior to electrode mounting the skin was shaved, sanded and cleaned with alcohol to reduce skin impedance. The electrodes were connected to lightweight preamplifiers equipped with an A/D-converter with 16 bit resolution. The signals were transmitted from the preamplifiers through wires to a recording box (MQ16, Marq Medical) where data were band-pass filtered (10-1000 Hz). The recording box transferred data wirelessly via Bluetooth-technology to a PC, where data was sampled using a custom-written Matlab-script. The quality of the signals was checked on the computer screen, where data were displayed in real-time. EMG was sampled at 512 Hz.

After the mounting of the electrodes, the maximal EMG amplitude (EMGmax) was measured during three isometric contractions for all muscles. For the anterior deltoid muscle the subject was standing with the right shoulder flexed 30 degrees. The measurement was performed while the subject pushed a tight nylon strap upwards with the back of the hand. The EMGmax recording for the intermediate deltoid was performed similarly, but with the shoulder in 30 degrees abduction. For the trapezius muscle, the subjects elevated the right shoulder against the resistance of a tight strap fixed to the floor. For both m. erector spinae parts the subjects extended the trunk against the resistance of a nylon strap around the shoulders, while the anterior part of the pelvis was supported against a plate (Potvin 1996).

The works tasks were performed as a natural part of the baggage handling job. If the baggage handler was assigned to a certain task, this was not altered due to the data recording.

EMG data analysis

The full day measurements were divided into task specific measurements based on trigger signals from the start and end of tasks. Out of the total 102 we had 27 tasks specific measurements (15 baggage compartment, 12 conveyer, 5 baggage hall) from the fullday measurements. Data analysis was performed by a custom Matlab-script. Both amplitude probability distribution functions (APDF) and rolling root mean square (RMS) amplitude were calculated. In both cases EMG-signals were band-pass filtered at 10-250 Hz using a fourth order Butterworth filter. The EMG signals were visually and manually inspected for unrealistic spikes, drift and short periods of high noise. These were rare and removed before further analysis.

The method described by Jonsson et al. (1982) was used to produce APDF curves. Also according to Jonsson et al. (1982), three levels of activity were selected for further analysis (Figure 2).

The 10th percentile (P10) was considered the static level, the 50th percentile (P50) was the median level, and the 90th percentile (P90) was considered the peak level of activity (Jensen 1993; Jonsson 1982).

Rolling RMS windows of one second (RMS1), 5 seconds (RMS5), and one minute (RMS60) were calculated and expressed relative to EMGmax (%EMGmax). The peak values from the three RMS analyses along with the P10, P50 and P90 from the APDF analysis were input to the statistical analysis.

Biomechanical loading analysis

Initially the biomechanical loading analysis was performed on all subtasks in the three work tasks. However, because it was impossible to isolate the EMG measurements in the single subtasks, we decided to collapse the biomechanical loading analysis into the same three more general tasks for comparability reasons. We therefore report on the results with both methods.

The compression force and flexor/extensor moment between the L4/L5 vertebrae and the right shoulder flexor moment were calculated for the same work tasks (baggage hall, baggage compartment and by the conveyor) as the EMG analysis. In each task the baggage handler was video recorded from a sagittal view. From the video five still images representing different parts of the handling task were extracted. Segment angles for foot, shank, thigh, torso, head, upper arm, forearm and hand were measured on the still images with ImageJ (National Institute of Health, USA). The segment angles were used as input to a nine segment rigid body Watbak model (University of Waterloo, Canada) which calculated the compression force and joint moment at L4/L5-level and shoulder flexor moment for the right arm. The Watbak model was a static 2D model. It included a single back extensor muscle with a moment arm of 6 cm that balanced the extensor reaction moment of the lumbar spine (Cholewicki 1991). The extensor muscle worked at an angle of 5 degrees to the direction of compression force (Cholewicki 1991).

For each of the five still pictures from every lift analysis 10 kg, 15 kg and 20 kg were used as baggage weight. In each of the subtasks 10 different baggage handlers were filmed. Some baggage handlers were filmed in several tasks, so a total of 44 individual baggage handlers were included. To make the results comparable, all biomechanical parameters are expressed relative to body mass.

Statistical analysis

A linear mixed model with post-hoc tukey-corrected multiple comparisons performed in SAS 9.3 (SAS institute Inc., Cary, NC, USA) was applied to identify statistically significant differences between the three tasks in spinal loading and levels of muscle activity. Level of significance was set to 5 %.

Results

EMG

Relative muscular activity for all APDF levels, muscles, and tasks are presented in Table 2. In all APDF activity levels and muscles (except for the erector spinae L4/L5, p10 and trapezius, p50) the baggage compartment task had the highest level of activity. This did not reach statistical significance. In the ADPF-analysis of the full day recordings (Table 3) all activity levels were equivalent to what was found in the task-based analysis (Table 2)

Table 4 contains peak levels of muscle activity from RMS1, RMS5, and RMS60. In the intermediate deltoid, the baggage compartment task had significantly higher muscle activity than the baggage hall task. No task had higher general level of muscle activity in the remaining muscles.

Muscle	**Deltoideus ant.**	**Deltoideus int.**	**Erec. Spin. L4/L5**	**Erec. Spin. Th12**	**Trapezius**
P10 (%EMGmax)					
Baggage hall	0.7 (0.2)	0.6 (0.4)	3.1 (1.0)	4.1 (1.1)	2.4 (0.4)
By conveyor	0.6 (0l2)	0.8 (0.3)	4.2 (1.0)	4.5 (1.2)	1.7 (0.4)
Baggage compartment	0.6 (0.2)	0.9 (0.3)	3.5 (0.7)	6.0 (0.8)	1.5 (2.9)
P50 (%EMGmax)					
Baggage hall	3.5 (1.4)	2.8 (1.1)	8.4 (2.7)	11.8 (3.3)	7.1 (1.0)
By conveyor	3.3 (1.5)	3.5 (1.0)	12.6 (2.7)	14.5 (3.5)	6.0 (1.0)
Baggage compartment	4.5 (1.0)	4.2 (0.8)	12.9 (2.0)	18.1 (2.4)	6.6 (0.8)
P90 (%EMGmax)					
Baggage hall	19.7 (4.3)	11.8 (3.6)	21.9 (5.1)	26.8 (7.2)	17.6 (2.7)
By conveyor	18.1 (4.0)	17.3 (3.3)	33.9 (5.6)	38.7 (7.7)	20.3 (2.9)
Baggage compartment	23.2 (3.1)	19.4 (2.6)	34.9 (3.9)	41.9 (5.3)	23.6 (2.2)

Table 2. APDF in five muscles and three tasks. Mean (SE)

	Deltoideus ant.	Deltoideus int.	Erec. Spin.L4	Erec. Spin. Th12	Trapezius
P10	0.9 (0.3)	0.3 (0.04)	2.5 (0.3)	4.8 (0.9)	3.8 (2.2)
P50	6.3 (2.2)	2.6 (0.5)	9.6 (1.4)	12.3 (1.5)	11.4 (4.6)
P90	23.8 (5.5)	19.8 (3.9)	41.2 (9.3)	28.2 (2.8)	29.4 (10.5)

Table 3. APDF based on full day recordings from five muscles, but not divided into tasks. Mean (SE)

Muscle	Deltoideus ant.	Deltoideus int.	Erec. Spin.L4	Erec. Spin. Th12	Trapezius
RMS1 (%EMGmax)					
Baggage hall	99.8 (14.8)	51.1 (6.6)†	90.0 (37.1)	100.3 (32.7)	63.3 (10.0)
By conveyor	69.6 (12.6)	64.5 (6.7)	96.1 (34.3)	104.7 (34.4)	72.2 (9.4)
Baggage compartment	80.6 (12.6)	77.5 (4.7)	113.4 (24.87)	123.5 (22.6)	63.8 (6.9)
RMS5 (%EMGmax)					
Baggage hall	66.5 (10.0)	30.1 (4.0)†	50.7 (23.6)	58.1 (21.0)	40.6 (5.9)
By conveyor	41.8 (8.6)	36.7 (4.2)	56.8 (22.4)	73.8 (21.8)	44.0 (5.7)
Baggage compartment	50.2 (6.6)	48.1 (2.9)	72.1 (16.0)	79.0 (14.4)	37.9 (4.1)
RMS60 (%EMGmax)					
Baggage hall	40.3 (7.7)	13.9 (2.4)†	24.8 (11.4)	34.8 (11.5)	20.7 (3.1)
By conveyor	20.4 (6.5)	18.1 (2.4)	34.2 (10.7)	44.4 (11.5)	23.5 (3.1)
Baggage compartment	25.6 (5.1)	24.6 (1.7)	40.0 (7.7)	44.7 (7.8)	20.5 (2.2)

Table 4. Rolling RMS averages in five muscles and three tasks. †: hall ≠ compartment indicate statistically significant differences at $p < 0.05$. Mean (SE)

Biomechanical loading

The L4/L5 extensor moments, compressions and shoulder moments from the general tasks are presented in Table 5 and estimates from the subtasks are presented in Table 6. The L4/L5 extensor moment in the baggage compartment task was significantly higher than in the two other tasks (Table 6). The compression force between L4 and L5 in the baggage compartment task was significantly higher than the conveyor task and the baggage hall task. There was no difference between the conveyor task and the baggage hall task (Table 5). The biomechanical variables increased significantly ($p<0.001$) with increasing baggage weight in all tasks.

There were no significant differences in the shoulder flexor moment between the tasks.

Task/Baggage weight	10 kg	15 kg	20 kg
Compression (N/BM)			
Baggage hall	22.6 (0.5)†	27.3 (0.6)†	32.0 (0.7)†
By conveyor	21.3 (0.6)◊	26.2 (0.7)◊	31.1 (0.8)◊
Baggage compartment	29.0 (1.0)	34.1 (1.1)	39.0 (1.3)
Extensor moment (Nm/BM)			
Baggage hall	0.96 (0.03)†	1.20 (0.04)†	1.44 (0.05)†
By conveyor	0.89 (0.03)◊	1.14 (0.03)◊	1.40 (0.04)◊
Baggage compartment	1.42 (0.07)	1.70 (0.08)	1.97 (0.08)
Shoulder moment (Nm/BM)			
Baggage hall	0.24 (0.01)	0.33 (0.01)	0.43 (0.02)
By conveyor	0.26 (0.01)	0.37 (0.02)	0.48 (0.02)
Baggage compartment	0.22 (0.01)	0.33 (0.01)	0.40 (0.03)

Table 5. Compression force and extensor moment at the L4/L5 joint along with shoulder flexor moment. All are relative to body mass. †: hall ≠ compartment, ◊: conveyor ≠ compartment indicate statistically significant differences at $p < 0.05$. Mean (SE)

Task/Baggage weight	10 kg	15 kg	20 kg
Compression (N/BM)			
Loading cart	20.9 (0.82)	25.5 (0.95)	30.1 (1.1)
Unloading cart	21.9 (0.75)	27.0 (0.89)	32.0 (1.0)
Stooped	42.0 (0.96)†	47.8 (1.2)†	53.9 (1.3)†
Kneeling	26.7 (0.96)	31.8 (1.1)	36.2 (1.2)
Sitting	18.4 (1.3)	22.7 (1.6)	27.0 (1.9)
Unloading container	22.8 (1.0)	26.6 (1.3)	30.3 (1.6)
Loading container	24.9 (1.3)	30.3 (1.6)	35.7 (1.8)
Extensor moment (Nm/BM)			
Loading cart	0.87 (0.05)	1.11 (0.06)	1.35 (0.07)
Unloading cart	0.91 (0.04)	1.18 (0.05)	1.45 (0.06)
Stooped	2.40 (0.05)†	2.74 (0.07)†	3.08 (0.07)†
Kneeling	1.23 (0.06)	1.51 (0.07)	1.73 (0.08)
Sitting	0.62 (0.09)	0.87 (0.10)	1.10 (0.12)
Unloading container	1.04 (0.07)	1.24 (0.08)	1.45 (0.09)
Loading container	1.02 (0.10)	1.27 (0.13)	1.52 (0.15)
Shoulder moment (Nm/BM)			
Loading cart	0.22 (0.01)	0.32 (0.02)	0.42 (0.02)
Unloading cart	0.29 (0.02)‡	0.41 (0.02)‡	0.54 (0.03)‡
Stooped	0.12 (0.03)	0.18 (0.04)	0.24 (0.05)
Kneeling	0.26 (0.02)	0.37 (0.03)	0.45 (0.04)
Sitting	0.30 (0.03)	0.40 (0.05)	0.51 (0.06)
Unloading container	0.12 (0.02)	0.16 (0.04)	0.19 (0.04)
Loading container	0.32 (0.02)*§	0.44 (0.02)*§	0.56 (0.03)*§

Table 6. Compression force and extensor moment at the L4/L5 joint along with shoulder flexor moment for each task. All are relative to body mass. †: Stooped ≠ all other tasks, ‡: unload cart ≠ unload container, §: unloading container ≠ stooped, *: unloading container ≠ sitting indicate statistical differences at $p < 0.05$. Mean (SE)

Discussion

We investigated the muscle activity and biomechanical loading of airport baggage handlers with the aim of elucidating the differences between the work tasks. The investigation was performed in a genuine work setting of the baggage handlers (Copenhagen Airport).

The only differences between the tasks in muscle activity were in the intermediate deltoid in the RMS analysis. However, the muscle activity was high in all tasks. The average peak RMS muscle activity exceeded 120 % EMGmax in the erector spinae Th12 during the baggage compartment task (Table 4). In the RMS analyses none of the tasks were consistently higher in muscle activity than the others, however in the APDF analyses the baggage compartment task was consistently demanding the highest muscle activity (Table 2). However, this was not statistically significant. Erector spinae had higher muscle activity levels at P10, P50 and P90 in all tasks than both the intermediate and anterior deltoid and the trapezius. Also this was not statistically significant. Regarding the biomechanical loading we found that the L4/L5 compression and extensor moment was significantly higher in the baggage compartment task than both the conveyor and baggage hall tasks. There was no difference in shoulder flexor moment.

The level of activity (APDF) in the trapezius was equivalent to the level of muscle activity in house painters in a laboratory setting (P10: 1.59 %EMGmax, P50: 6.8 % EMGmax, P90: 17.47 %EMGmax) (Meyland 2014). However, this was performed as intensive periods of work in different tasks as opposed to the present study that was performed in a genuine work setting where unexpected breaks in the tasks occurred. This may also be the reason for the lack of statistical differences between the full day recordings and the task based results. We expected that the task based results would show a higher level of activity than the full day recordings, because all breaks and other types of less strenuous work tasks were included. However, the APDF analysis does not take the lengths of breaks into account. So a baggage handler performing the conveyor task could have several small periods without baggage handling, and the results from the APDF would probably be similar to those from a baggage handler who had a long break and then more continuous strenuous work. This means that we may have underestimated the muscle activity in the work tasks of the baggage handlers because the work task did not solely consist of the work task but involved a lot of small breaks also. However, the results do reflect the actual activity demands, as the recordings were done in the genuine work environment of the baggage handlers.

Furthermore, the comparison between the three tasks may not be valid because two tasks (conveyor and baggage compartment tasks) were characterized by short bouts of hard work followed by a break and the third task (baggage hall) was characterized by more continuous and less intensive work. These differences between the tasks were not quantified, but in general the baggage handler in the baggage hall would handle all the baggage for a flight in two hours (from check-in opens two hours before departure), whereas the baggage handlers on the ramp had only about 20-30 minutes. This was probably the reason why the baggage hall task had the lowest level of activity in P50 and P90 (Table 2). However, it was surprising that there was no significant difference between the baggage hall task and the two other tasks.

Jonsson (1982) suggested a set of limits for P10, P50 and P90 muscular activity level. Activity should not exceed 2 %EMGmax and must not exceed 5 %EMGmax for the static level (P10), 10 %EMGmax should not be exceeded and 14 %EMGmax must not be exceeded for the median level (P50), and 50 %EMGmax should not be exceeded and 70 %EMGmax must not be exceeded for the peak level (P90). The static activity level in the erector spinae L4/L5 exceeded the 2 % level, but not 5 % (Table 2). Furthermore, the baggage compartment task in the erector spinae Th12 exceeded the 5 % limit, whereas the remaining two tasks exceeded the 2 % limit (Table 2). In addition the baggage compartment task in the trapezius exceeded the 2 % limit. The P50 limit of 10 % was exceeded in all tasks except the baggage hall task. In addition, the 15 % limit was exceeded in the baggage compartment task in the erector spinae Th12. None of the tasks exceeded the P90 limits. This was surprising because the baggage handler lifts suitcases with a mean weight of 15 kg, and accumulates 5 tons per day on average (Brauer 2013). Therefore we expected that the level of muscle activity would be higher than what we found. This is even lower than what has been found in analysis of seamstresses (Jensen

1993) and persons working at a pillar drill (Christensen 1986). The low static level in the present study can probably be explained by the lack of static work in baggage handlers. The nature of the tasks was almost purely dynamic. Therefore, the P10 probably expressed the activity during the resting periods of the tasks. However, this does not explain the low median and peak values. On the contrary we expected the P90 to be much higher, as the tasks involved heavy lifting in awkward positions. This may be caused by issues in the data collection process. Sometimes the baggage handlers would wait for several minutes during the recording for baggage to arrive to the aircraft. This may have caused us to underestimate the level of muscle activity.

The RMS analysis showed some large muscle activity levels exceeding 100 %EMGmax. This reflects the protocol used to measure EMGmax. The EMGmax contractions were isometric, whereas the work tasks were dynamic, hence including anisometric contractions and muscle contractions at varying velocities and muscle lengths.

In a study of dentists Finsen & Christensen (1998) found that a max level of 17 %EMGmax in m. trapezius during cavity filling with a one second rolling RMS window. In comparison we found 72 %EMGmax on average for baggage handler tasks (Table 4). This is not surprising since the work as a baggage handler is obviously more strenuous than dentist work.

In the biomechanical loading analysis we found that the level of compression in the L4/L5 segment did not exceed the NIOSH recommendations of 3400 N for the average baggage handler (82.6 kg) (Bern 2013) in any of the general tasks. One explanation for the low level of compression force in the baggage compartment task is that this was an average of several positions including kneeling, stooped, and sitting. In the stooped task we recorded larger compression forces (4460 N), whereas the sitting task only produced around 2230 N of compression (Table 6). This is not an unreasonable conclusion, as the baggage handler can switch between positions at will. A drawback to the Watbak-models is that they are very simple. They are two-dimensional and static, thus they do not take into account the movements in other than sagittal direction, nor the accelerations of the body and burden that is handled. This will most likely underestimate the compression forces and joint moments.

Moreover, the model only contains one muscle producing the lumbar extensor moment with a fixed moment arm of 6 cm. This is a very crude assumption since there are many muscles balancing the extensor moment and they originate and insert at different sites, thus producing force on the lumbar spine with individually different moment arms that vary with body size.

In conclusion, the baggage handlers all worked in their usual environment, and they were specifically asked to perform their job as usual and not take special notice to the measurements. Therefore all circumstances (i.e. time pressure, weather, baggage delay) were all in accordance with the conditions of the baggage handlers. The levels of muscular activity exceeded 100 %EMGmax in the RMS analysis. Even though there were high peaks of muscle activity we have probably underestimated the isolated activity demands of the work tasks. However, the results do present a valid estimate of the general level of muscle activity of the baggage handler. There were very few statistically significant differences between the tasks in the muscular activity. However, the baggage compartment task put significantly more load on the lumbar spine than the other tasks. In the future this work can potentially be used to prioritize rotation between work tasks.

Acknowledgements

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Conflicts of interest

There have been no conflicts of interest

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Paper II

The validation of a musculoskeletal model of the lumbar spine

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Abstract

Musculoskeletal models are the only indirect method for investigating spinal forces. However, the validation of musculoskeletal models is difficult because of the challenge of obtaining muscle- and joint forces from in vivo studies. The purpose of the present study was to compare the compression forces obtained by the AnyBody Modeling System lumbar spine model lumbar spine model to the experimentally measured in vivo intradiscal pressures published by Wilke et al. in 2001

Inverse dynamic-based models were built using the Anybody Modeling System v. 6.0.4. The models were scaled to the bodily measures provided in the paper by Wilke et al., and placed in nine different positions that matched the in vivo measurements. The intradiscal pressures from Wilke et al. were converted to compression force. Compression forces with two different muscle recruitment criteria were estimated by the model and compared to the converted in vivo pressure measurements.

When a 2nd order polynomial criterion was applied the agreement between the measured and the estimated L4/L5 compression forces was very high and errors nearly negligible. Especially for high levels of spinal forces, the differences between measured and estimated compression forces were negligible (< 10 %). In addition, the model responded adequately to different positions regardless of muscle recruitment criterion.

Keywords:

Validation; Musculoskeletal model; Compression; Modeling; Spine

Introduction

In 2007, de Zee et al. (2007) published a new highly detailed model of the lumbar spine mainly based on the review of the spinal anatomy by Hansen et al. (2006). This model is available in the AnyBody Managed Model Repository (AMMR) accompanying the AnyBody Modeling System (AMS) (Damsgaard et al., 2006) and comprises segments, more than 190 muscle fascicles, abdominal muscles and a model of the intra-abdominal pressure. The model is based on inverse dynamics and can output muscle forces, joint forces and joint moments. The compression forces in the spine are important, because back pain has a high prevalence in occupations with high levels of spinal compression forces (Bern et al., 2013; Burdorf, 1992; Krause et al., 2004; Latza et al., 2000; Seidler et al., 2009). Therefore, it is desirable to be able to estimate the compression forces validly without the use of invasive techniques. Musculoskeletal models are the only indirect method for investigating spinal forces. However, the validation of musculoskeletal models is difficult due to the challenge of obtaining muscle- and joint forces from in vivo studies. Spinal models can be particularly difficult to validate, because intra-discal pressures can only be recorded by invasive methods. A few studies have measured spinal pressure during standing, sitting and lifting. The first reports were published by Nachemson (Nachemson, 1966b; Nachemson, 1966a; Nachemson, 1965; Nachemson and Morris, 1964) and later repeated and refined by others (Wang et al., 2014; Lisi et al., 2006; Ledet et al., 2005; Wilke et al., 2001; Sato et al., 1999; Wilke et al., 1999). In 1999, Wilke et al. (1999) published the intra-discal pressure from a single subject during sitting and standing postures and different lifting tasks. This study was followed by another (Wilke et al., 2001) in which the authors provided details of the study for validation purposes. Validation of absolute values as well as validation of the model's response to changes is important in musculoskeletal models (Lund et al., 2012). Obviously, absolute validation is of predictive interest, but it is also crucial that the model parameters interact correctly, so that the model responds adequately to changes. Musculoskeletal models must be validated to the purpose of use. The model can only predict the size of the compression force if it provides valid absolute output. However, the model can still be used to investigate differences between positions or tasks provided that the model responds adequately to the changes in conditions (Lund et al., 2012). The comparison of spinal forces from AMS to intra-discal pressure has been investigated by Rajaee et al. (2015) but they did not consider the effects of changes in task or position or effects of different muscle recruitment criteria. Arjmand et al. (2011) have also made this type of validation with the Wilke et al.-data set using another spine model than the AMS.

The purpose of the present study was to compare the compression forces obtained by the AMS lumbar spine model to the experimentally measured in vivo intra-discal pressures (Wilke et al., 2001; Wilke et al., 1999). Both validation of absolute forces and responsiveness to changes were emphasized. Furthermore, we wanted to investigate which muscle recruitment criterion was the most appropriate.

Methods

Inverse dynamic-based models were built using AMS v. 6.0.4 (Damsgaard et al., 2006). The models were all modifications of the "StandingModel", which is freely available in the AMMR v. 1.6.2. The base model was scaled to fit the bodily measures of the subject in the Wilke et al.-study (Wilke et al., 2001) (72 kg, 173.9 m). Segment masses and lengths were scaled according to Winter et al (2005).

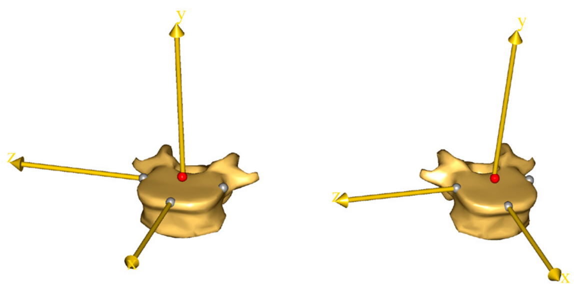


Figure 1. The anatomical reference frame. Compression force is measured in the Y-direction.

The output parameter (compression force) was measured in local coordinates on the cranial endplate of the L5 vertebra. The L5 endplate formed a plane to which the compression force was perpendicular (Figure 1).

The muscle redundancy problem was solved with two different criteria: 1) by minimizing the sum of muscle activities squared (2nd order polynomial) and 2) according to a minimum fatigue criterion (min/max criterion). The latter relies on the assumption that muscles are recruited in a manner to postpone fatigue (Rasmussen et al., 2001) and is seeking to activate all synergists with the least possible relative activation. Results from the two muscle recruitment solvers are reported separately.

The positions that were analysed are depicted in Figure 2. We compared common positions in daily living (lying, sitting, standing, standing flexed) adapted from Wilke et al. (2001), and since descriptions of velocities and accelerations were not provided by Wilke et al. (2001), we chose to analyse the positions that were static or involved static lifting only. In the positions where the model is lying or seated, the connection between the human model and table or chair was modelled using conditional contact elements.

This contact model was similar to the one published by Rasmussen et al. (2009). These conditional contacts enable computation of the reaction forces between the model and the environment as part of the muscle recruitment.

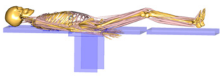
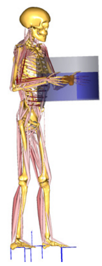
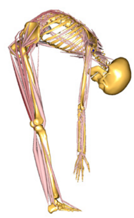
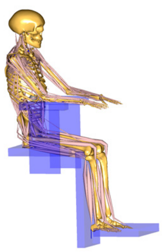
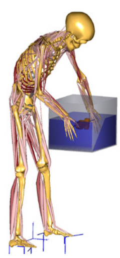
	Lying supine on a table.		Standing flexed The back in 60 degrees flexion. Arms vertical. 50 and 70 degrees of flexionare also presented.
	Sitting relaxed The back was extended 10 degrees, arm resting on armrests.		Lifting a box close to the body. The back straight and the box close to the chest.
	Standing straight The back was extended 5 degrees.		Maximum flexion The back in 95 degree flexion, hip in 50 degree flexion. Arms pointing towards toes.
	Sitting straight The back in 10 degree flexion, with no support of the back but arms resting on the armrests.		Lifting a box with stretched arms Back straight, shoulders 60 degrees flexion.
		Lifting a box with flexed back The back in 60 degrees flexion. The box 70 cm above the floor. 50 and 70 degrees of flexion are also presented.	

Figure 2. Nine different positions of the model in Paper II.

When the model was standing, the conditional contacts simulated the ground reaction forces. This method for estimation of ground reaction forces has been validated by Fluit et al. (2014). During lifting the contact between the box and the hands was modelled as a spherical joint on each hand. This was balanced by five virtual muscles on each hand. Each muscle had a strength of 4000 N. The box had a mass of 20 kg. In the “Standing flexed” and “Lifting with a flexed back” models, the flexion between the pelvis and the thorax was set to 60 degrees. This was an estimate based on photographs from the two papers by Wilke et al (Wilke et al., 2001; Wilke et al., 1999). To investigate the model’s sensitivity to the flexion angle and to assess the uncertainty of position estimation, we also analysed the compression force in 50 and 70 degrees of flexion in both positions.

In order to compare the in vivo measurements from Wilke et al. (2001) with the compression forces from the models, the spinal pressures (MPa) were converted to force (N) by:

$$F=PAC_{corr,}$$

where P is the measured intra-discal pressure, A is the cross-sectional area of the L4/L5 intervertebral disc (1800 mm2) obtained from an MRI scanning and reported along with the pressure measurements (Wilke et al., 2001) and C_{corr} is a correction constant of 0.77. This correction factor compensates for the heterogeneous material composition and therefore non-uniform loading of the disc. The correction factor has shown good correlation between intra-discal pressure and compression force in a finite element model for all body positions except standing with an extended back (Dreischarf et al., 2013).

The primary outcome measure in the present study was the compression force between L4 and L5 from the models. We examined the absolute and relative differences between measured and predicted compression forces in nine different body positions. The validity was assessed as the average difference and as the trend of differences across measured compression forces. The compression forces based on two muscle recruitment algorithms along with the converted spinal pressures from Wilke et al. (2001) are presented.

Results

The measured and estimated compression forces are depicted in Figure 3. The estimated compression forces and their differences from the measured compressions are shown in Table 1.

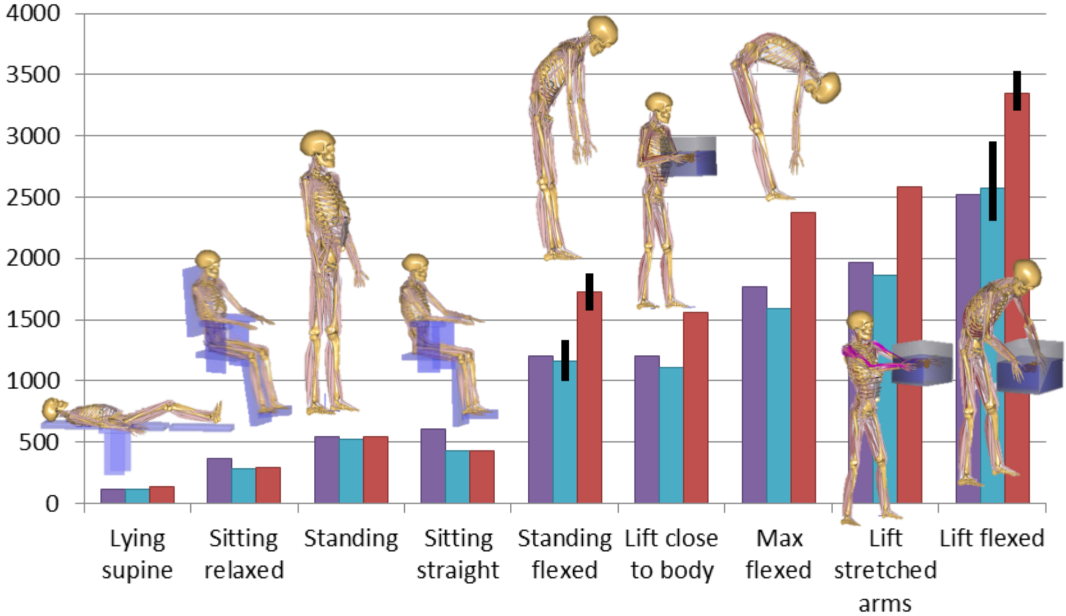


Figure 3. Estimated compression forces from the model and in vivo measurements. Purple: in vivo measurements, turquise: 2nd order polynomial, red: min/max criterion. Black bars represent compression forces in 50 and 70 degrees of flexion.

When the 2nd order polynomial criterion for muscle recruitment was applied there was high agreement between the experimental and the modelled results. The largest absolute error was in the "sitting straight" and the "max flexed"-positions and was 176 N (resp. 29 % and 10 %) lower than in vivo data. The average relative error was -9% with the 2nd order polynomial and 16 % with the min/max criterion. When the in vivo compression force exceeded 1200 N, the min/max criterion generally overestimated the forces, whereas the 2nd order polynomial predicted the forces accurately with errors below 11 %. The largest absolute error with the min/max criterion was 831 N (33 %) in "lifting with flexed back" (Table 1). With measured values exceeding 1200 N the average error for the 2nd order polynomial was -5 % and 34 % with the min/max criterion.

When the compression forces were low both recruitment criteria produced comparable results, and regardless of muscle recruitment criterion the model predicted the changes in spinal compression well (Figure 3).

Position/ Measurement	Wilke in vivo (N)	2nd order polynomial (N)	Difference (N / %)	Min/Max-criterium (N)	Difference (N / %)
Lying supine	110	113	3 / 3	138	28 / 25
Sitting relaxed	361	281	-80 / -22	290	-71 / -20
Standing	548	518	-30 / -5	548	0 / 0
Sitting straight	602	426	-176 / -29	424	-178 / -30
Standing flexed (60°)	1205	1159	-46 / -4	1730	525 / 49
Lift close to body	1205	1104	-101 / -8	1553	348 / 29
Max flexed	1766	1590	-176 / -10	2375	609 / 34
Lift stretched arms	1971	1862	-109 / -6	2581	610 / 31
Lift flexed back (60°)	2519	2573	54 / 2	3350	831 / 33

Table 8. Absolute compression forces from two muscle recruitment criterions and the in vivo study. Error is the difference between the modeled estimate and the in vivo measurement.

Discussion

We have presented a comparison between L4/L5 intra-discal pressures measured in vivo and estimates of L4/L5 compression force from a musculoskeletal model with two different muscle recruitment criteria. When the 2nd order polynomial criterion was applied the agreement between the measured and the estimated L4/L5 compression forces was very high and errors nearly negligible (Table 1). Especially for high levels of spinal forces the relative differences between measured and estimated compression forces were negligible (< 10 %).

To be able to compare different positions and investigate differences in compression force, the model must be sensitive to changes in compression force between positions and tasks. In the present study, the model showed high sensitivity to the compression force between positions and a high degree of agreement with the changes in the measured intra-discal pressure. Even though the absolute errors with the min/max criterion were large, the response to changes in conditions was adequate. Even when the forces were low, the model predicted the change in the measured compression between positions fairly well.

The present validation study on the spine model shows that the 2nd order polynomial for muscle recruitment is a more appropriate recruitment criterion than the min/max criterion when the muscle forces larger than 1200 N. When the muscle forces are low the min/max and 2nd order polynomial produce the same level of compression.

There are some limiting factors to this validation study. The positions of the model were estimated based on descriptions and photographs from Wilke et al (2001). The validity of the estimations would have improved markedly if kinematic data or segment angles had been available. Also, segment properties were estimated based on the anthropometric fractions by Winter(2005). Wilke et al.(2001) did report on a variety of anthropometric parameters, but these were not applicable with the required segment lengths in AMS. Furthermore, the conversion of pressure to compression force could be considered questionable. Even though Dreischarf el at. (2013) used the correction factor of 0.77 to produce good estimations of force; the correction factor was model specific. Hence, it is unknown if an identical correction factor should be applied to this model. To elucidate the validity of the correction factor and the conversion from pressure to force, a finite element analysis investigating the tissue-response to different types of compression in this specific model should be conducted. However, this is beyond the scope of the present study. We chose to use the correction factor, because it is the best according to present knowledge. Furthermore, a simple conversion without a correction factor as seen in previous studies (Nachemson and Morris, 1964; Nachemson, 1965; Nachemson, 1966a; Nachemson, 1966b; Sato et al., 1999) could result in large overestimations of the compression forces (Dreischarf et al., 2013). In general inverse dynamic models neglect the possible effect of co-activation of antagonistic muscles. In the present model, the muscle recruitment solver may co-activate muscles on both sides of a joint if there is need for a punctum fixum to solve the kinematics.

In conclusion, the musculoskeletal model with a 2nd order polynomial muscle recruitment criterion showed good absolute agreement with in vivo measured intra-discal pressures. Moreover, the trend agreement with the measured pressures was solid. The 2nd order polynomial criterion for muscle recruitment performed better than the min/max criterion for compression levels above 1200 N, and it is therefore recommended to use this criterion when estimating spinal compression forces up to 2500 N. However, there is no indication that the agreement will cease with increasing forces.

Conflict of interest

John Rasmussen and Mark de Zee are co-founders of the company, AnyBody Technology A/S, owning the AnyBody Modeling System, which was used for the simulations. John Rasmussen and Mark de Zee are minority shareholders on the company. John Rasmussen furthermore serves on the board of directors of the company.

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Paper III

Spinal loads in asymmetrical and dynamic lifting tasks: A modeling approach

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Abstract and Keywords

The purpose of the study was to estimate lumbar forces at L4/L5 during two heavy lifting tasks using a detailed musculoskeletal model and to compare this with vertebral compression tolerance and recommended lifting limits.
We built an inverse dynamics-based musculoskeletal computer model using the AnyBody Modeling System. 3D motion capture recorded a stooped and a kneeling lifting task simulating airport baggage handler work. Marker trajectories were used to drive the model. AnyBody-models estimated compression forces, shear forces and moments at the L4/L5 joint. Muscle activities from the back- and abdominal muscles were extracted.
The stooped lifting task produced 5541 N compression in the L4/L5 joint and a kneeling task produced 4197 N. These compression forces were close to the average compression tolerance and exceed the recommended limits.
The estimated peak compression forces during baggage handling tasks indicate that lumbar spine compression tolerance and recommended lifting limits may regularly be exceeded.

Key words:
Low back pain; lumbar loading; spinal compression; micro fractures; compression tolerance; lifting; asymmetrical; computer modeling; motion capture; lifting recommendations.

1.0 Introduction

Low back pain constitutes a major public health and economic challenge in many countries. According to the World Health Organization, low back pain is the seventh largest burden of disease in the world. In the Global Burden of Disease 2010 study low back pain accounted for 3.3% of total disability-adjusted life years lost, globally ranking 6th among all causes, and in regions of developed countries as 1st to 3rd cause (1). The causes of low back pain are still largely unknown. Several risk factors have been suggested including cigarette smoking (2), high psychological work pressure (3), previous episodes of low back pain (4), and heavy lifting (5-7). Heavy lifting causes high compression forces of the tissues in the low back (8;9). Micro-fractures in the endplates of the vertebrae have been suggested as a source of unspecific pain (10) , and it is assumed that micro-fractures may result from high compression forces.

The National Institute of Occupational Safety and Health (NIOSH) in USA recommended a limit of 3400 N as the maximal compression force in the low back allowed during manual continuous work. This recommendation was based on computations on a two-dimensional static model of lifting (11). Despite experiments of material properties of vertebrae from human cadaver specimens, the critical levels of compression forces hazardous to human vertebrae are still disputed. Granhed et al. (12) reported compression tolerance between 810 and 10090 N among 21 cadavers. Moreover, 23 % of the specimens were damaged by forces lower than 3400 N. The critical levels of compression tolerance were closely related to bone density (12).

It is considered a drawback of the NIOSH recommended limits that they were computed based on a model which was too simple (13); a two-dimensional and static link-segment model does not take the influence of inertia or asymmetric lifting into account, nor does it account for the consequences of lifting with a straight or a bent spinal column. Thus, it is desirable to analyze lifting by more realistic models capable of handling dynamic and asymmetric conditions.

A highly detailed computer model of the human spine based on commercially available software (Anybody Technology, Aalborg, Denmark) was introduced by de Zee et al. in 2007 (14). This model included seven rigid segments (the pelvis, five lumbar segments and a thoracic segment), more than 150 muscle fascicles and a model of the intra-abdominal pressure (IAP). This model can quantify the loadings on the lumbar spine during dynamic and complex lifting situations i.e. one handed lifting and/or lifting with flexed and rotated trunk. From this model, it is possible to estimate joint reaction forces from all joints along with muscle activity and forces from all muscle fascicles in the model. In a previous study (15), the output of the model has shown high agreement with in vivo measurements of intra-discal pressure (16). In addition, Rajaee et al. (17) evaluated the Anybody Modeling System (AMS) spine model in 22 different positions and found good agreement with intra-discal pressures. We, therefore, consider this model a reliable tool for simulation of compression forces during lifting.

The purpose of the present study was to estimate lumbar load from this model during a stooped and a kneeling lifting task as performed by airport baggage handlers, and to compare these forces with previously reported vertebral compression tolerance data obtained in vitro (13) and to the recommended compression limits (11;13;18). Even though estimation of spinal forces has been addressed numerous times with many different approaches, the novelty in this study is that the estimates are performed using a model with a level of detail that has not been seen until now. We performed this study as a computer modeling study based on lab recordings of one subject performing the work tasks.

This study was performed as a part of the Danish Airport Cohort study (19). The general purpose of this study was to map and analyse musculoskeletal disorders in airport baggage handlers. This was done through an epidemiological part (19) and a biomechanical part. The idea was to add the biomechanically estimated musculoskeletal loading as an exposure to a cohort of baggage handlers. The work in this paper represents parts of the biomechanical exposure matrix.

2.0 Materials and Methods

The average age and self-reported height and weight of baggage handlers at Copenhagen Airport were retrieved from Bern et al. (19) and we recruited a male subject with these average characteristics (48 years , 87 kg, 1.81 m). He had 16 years of seniority as a baggage handler and was employed by a large ground handling company at Copenhagen Airport. As a part of the Danish Airport Cohort study, we conducted a series of two-dimensional static lifting analyses in baggage handlers (Koblauch et al. under review), and the subject chosen to constitute the model of the present study fell within 2 SD of all subjects regarding L4/L5 compression forces. He gave his written informed consent before participating in the study, which was approved by the local ethics committee (J.nr. H-3-2011-140).
Based on two weeks of observations of baggage handling at Copenhagen Airport and interviews with present baggage handlers, we selected 20 different work tasks that characterized most of a workday for a baggage handler. The 20 tasks were selected based on a time aspect or a physical loading aspect. A task should take up a considerable part of a workday or carry high work load. Out of the 20 tasks, we selected two tasks that were common to the baggage handlers, but unusual to other professions: Baggage handling in a kneeling position and in a stooped position. These tasks are commonly used to handle baggage inside the aircraft luggage compartment because of the limited space available. In addition, these lifting tasks have never been examined with a highly detailed model as in the present study.

2.1 Kneeling position

In general, the subject was instructed to handle a suitcase as if it was in the real airport setting. A certain speed was not specified, but a trial was considered successful if the subject approved that it was similar to lifts in the airport. The subject moved a standard suitcase (57x23.5x37 cm) from the floor using both hands and transferred it to the left and placed it on a platform 30 cm above the floor. Starting position was with the suitcase placed to the right of the subject at a 45° angle. The subject was instructed to transfer the suitcase to the designated destination at a 45° angle to the left (Figure 1). This lifting technique is frequently used by baggage handlers inside the aircraft luggage compartment lifting suitcases from the floor to a belt conveyer or vice versa.

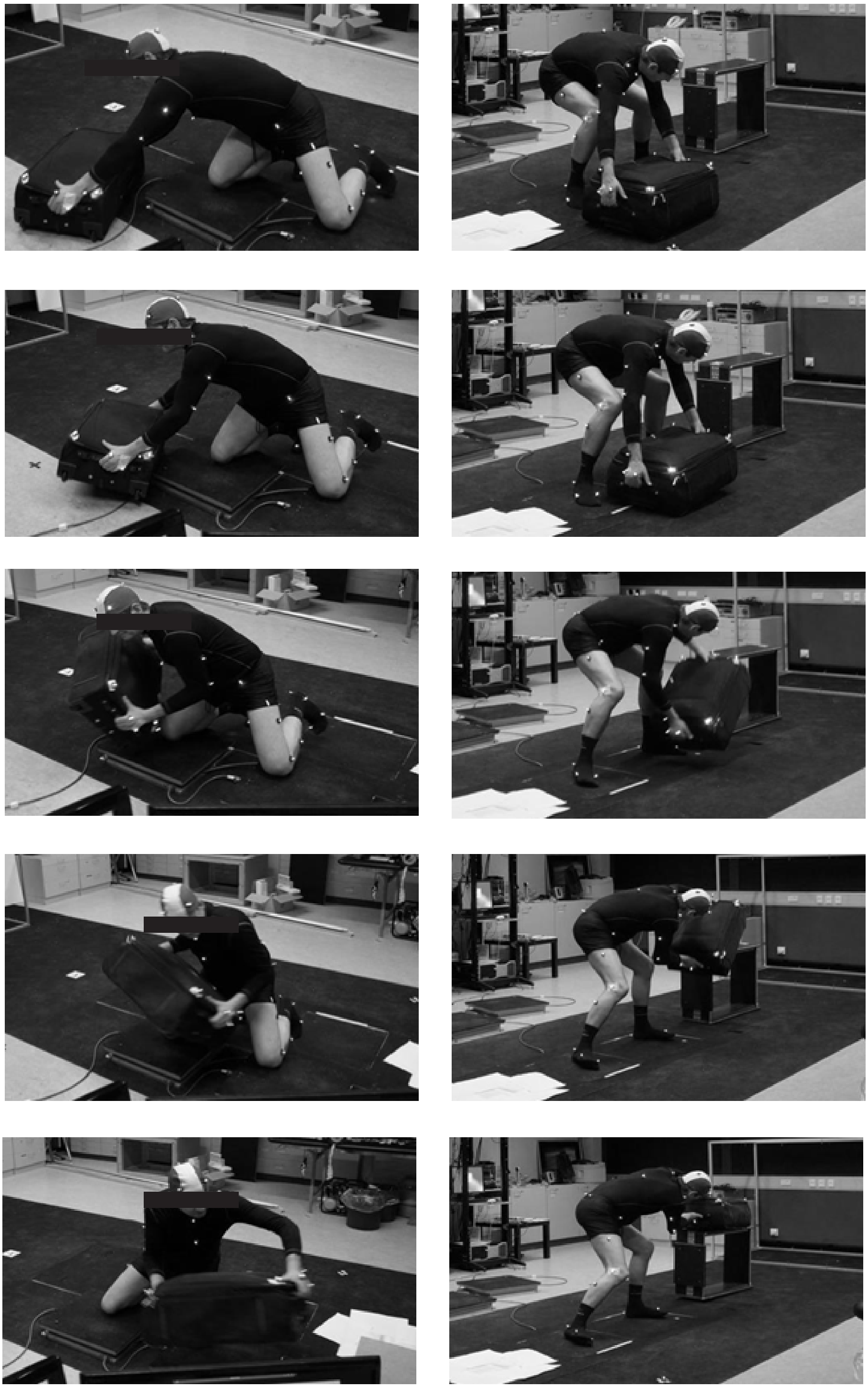

Figure 1. Time series of the two lifting tasks. Left: Kneeling. Right: Stooped

2.2 Stooped position

The subject was instructed to stand stooped but was allowed to bend his knees. The subject picked up the suitcase from the floor on the right side at a 20° angle using both hands and transferred it to the left in front of the body and placed it on a platform 50 cm above the floor. The platform was placed next to the subject at a 90° angle (Figure 1). This lifting technique is another option for baggage handlers inside the aircraft luggage compartment. However, this technique requires a higher ceiling in the aircraft than the kneeling position. This is also why the platform height was 50 cm and not 30 cm as in the kneeling task.

Three suitcase weights of 10 kg, 15 kg and 20 kg were used and both lifting tasks were performed experimentally in a laboratory. In the analysis, one trial from each task was used.

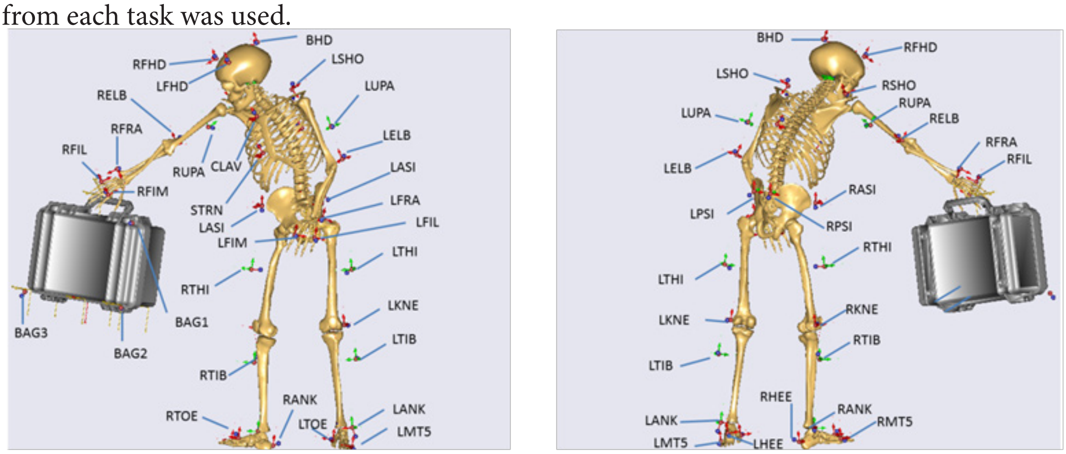

Figure 2. The marker setup used in the study

The two tasks were filmed at 75 frames per second by a custom-built motion capture system of eight synchronized high speed HD cameras (GZL-CL-41C6M-C, Gazelle, Point Grey, Richmond, Canada). The subject was equipped with a full-body marker setup of 37 luminous markers with a diameter of 5 mm while three markers were placed on the suitcase (Figure 2).

Two force platforms (AMTI, Watertown, MA, USA) measured ground reaction forces in the standing task, while four force plates were used in the kneeling task, one under each foot and one under each knee. Signals from all of the force platforms were sampled at 1000 Hz and were synchronized with the cameras. Three dimensional coordinates were computed with the Ariel Performance Analysis System (APAS, Ariel Dynamics Inc., San Diego, CA, USA) and were lowpass filtered at 6 Hz by a zero-phase lag, 4th order Butterworth filter. Inverse dynamics-based musculoskeletal models of the tasks were built in the AnyBody Modeling system v. 5.3 (AnyBody Technology A/S, Aalborg, Denmark). The models were modifications of the GaitFullBody model available from the AnyBody Managed Model Repository v. 1.5 (20) and were scaled to match the bodily measures of the subject through optimization using the method of Andersen et al. (21).
The highly detailed low back model including a model of the IAP described by de Zee et al. (14) was included in this full body model. In this model, joint reaction forces and torques were calculated by inverse dynamics, in which external forces and inertial properties of each segment are accounted for. The muscle activities were estimated according to a 2nd order polynomial optimization criterion. This criterion seeks to minimize the sum of muscle forces squared (22;23). This criterion proved superior in a previous validation of the lumbar spine model, where it was compared with another muscle recruitment criterion (min/max) (15). The six residual forces and moments were defined between the thorax and the ground. We estimated kinetics in the local L5 coordinates by defining three points on the cranial endplate of the L5-vertebrae forming a plane (Figure 3). In this plane, the z-axis was oriented in a medio-lateral direction between two lateral points. The x-axis was perpendicular to the z-axis in the endplate-plane and directed towards the anterior point on the L5. The y-axis was perpendicular to the z- and x-axis and the endplate-plane, originating from

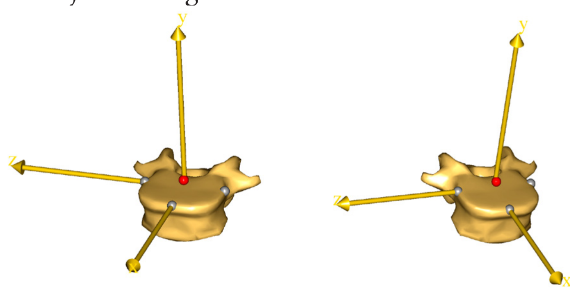

Figure 1. The anatomical reference frame. Compression force is measured in the Y-direction.

where the x-axis intercepted the z-axis. This was in accordance with the validation study by Koblauch et al. (15).

Furthermore, a suitcase-segment was added, which had the same spatial and inertial properties as the suitcase in the data collection. The model's right hand was linked to the suitcase by a revolute joint. The remaining degree of freedom was balanced by a dynamic contact model on the opposite end of the bag consisting of two contact points on the left hand and a cylindrical contact zone on the suitcase. Whenever the contact points were within the contact zone, a set of virtual muscles provided normal and frictional forces to balance the remaining degree of freedom, kinetically. This method was validated by Fluit et al. (24) for the prediction of ground reaction forces during activities of daily living. The activity of these virtual muscles was computed together with the remaining muscles in the muscle recruitment.

The peak, median and interquartile range (IQR) L4/L5-level compression, anterior/posterior (A-P) shear forces, and internal/external rotator moment were extracted from the musculoskeletal models for each of the three suitcase weight conditions. We also present the average muscle forces from the muscle fascicles of mm. erector spinae and mm. multifidi that passed the L4/L5 joint, and mm. obliquus interna et externa bilaterally.

3.0 Results

The spinal forces are presented in Table 1. For the 20 kg suitcase, the largest compression force was found in the stooped position (4692 N) and the largest A-P shear force (289 N) also in the stooped position. For the 15 kg suitcase, the largest compression force (4801 N) and A-P shear force (488 N) was also found in the stooped position. For the 10 kg suitcase the largest compression force (5541 N) and the largest A-P shear force (346 N) were found in the stooped position as well. In the stooped position, a peak of compression force occurred in the beginning of the task when the suitcase was accelerated (Figure 4).

Task	Weight (Kg)	Compression (N) (peak/median/ IQR)	Shear (N) (peak/ median/IQR)	Rotator moment (Nm) (peak/ median/IQR)
Kneeling	20	4197/2977/1051	237/148/71	69/9/79
Stooped	20	4692/3407/605	389/151/85	165/94/60
Kneeling	15	3341/2688/997	168/102/52	66/-2/75
Stooped	15	4801/3030/987	488/68/132	152/82/74
Kneeling	10	3039/2108/1067	125/98/70	47/-22/66
Stooped	10	5541/2740/3525	346/111/284	173/81/31

Table 1. The peak, median and inter quartile range for compression, A/P shear forces, and internal/ external rotator moment for 10 kg, 15 kg and 20 kg suitcase in the two tasks.

The largest peak of both compression and A-P shear forces occurred halfway through the task. This coincided with the instant at which the box was lifted off the floor. The peak compression and A-P shear forces in the kneeling position occurred in the last third of the task, where the subject lifted the suitcase towards his chest (Figure 5).

The maximal muscle force was 362 N in the right obliquus internus in the stooped position (Figure 4) and 135 N in the right obliquus externus in the kneeling position (Figure 5). In the stooped position, the first overall peak of muscle force coincided with the first peak in the compression and A-P shear force. Furthermore, the second peak of the left and right obliquus internus coincided with the largest peak of the compression force and A-P shear force (Figure 4). At the time of the overall peak of compression force, the right obliquus internus also showed a peak of force.

In the kneeling position, the peak of the right obliquus internus force occurred at the same instant as the largest peak of compression force (Figure 5).

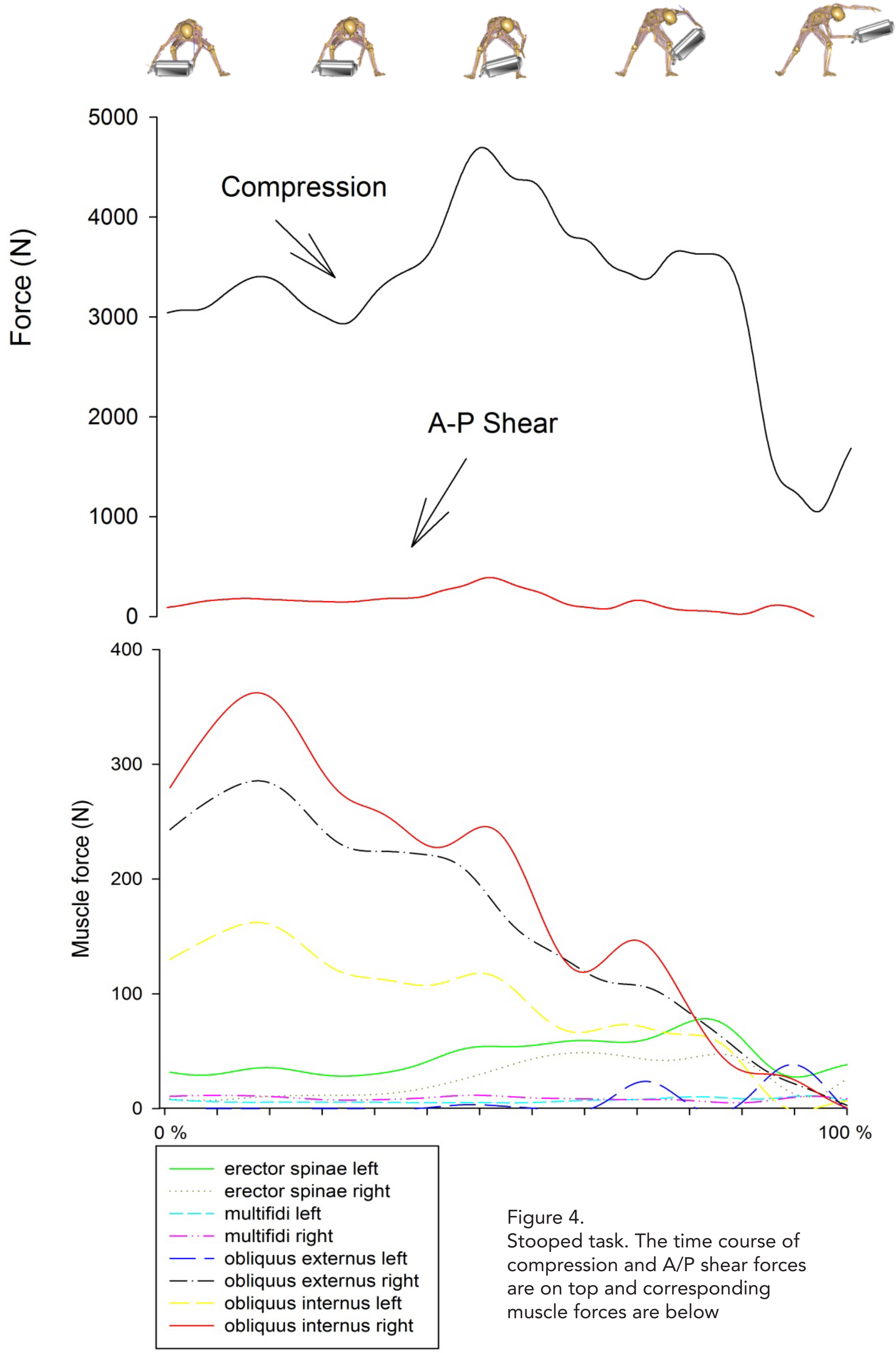

Figure 4.
Stooped task. The time course of compression and A/P shear forces are on top and corresponding muscle forces are below

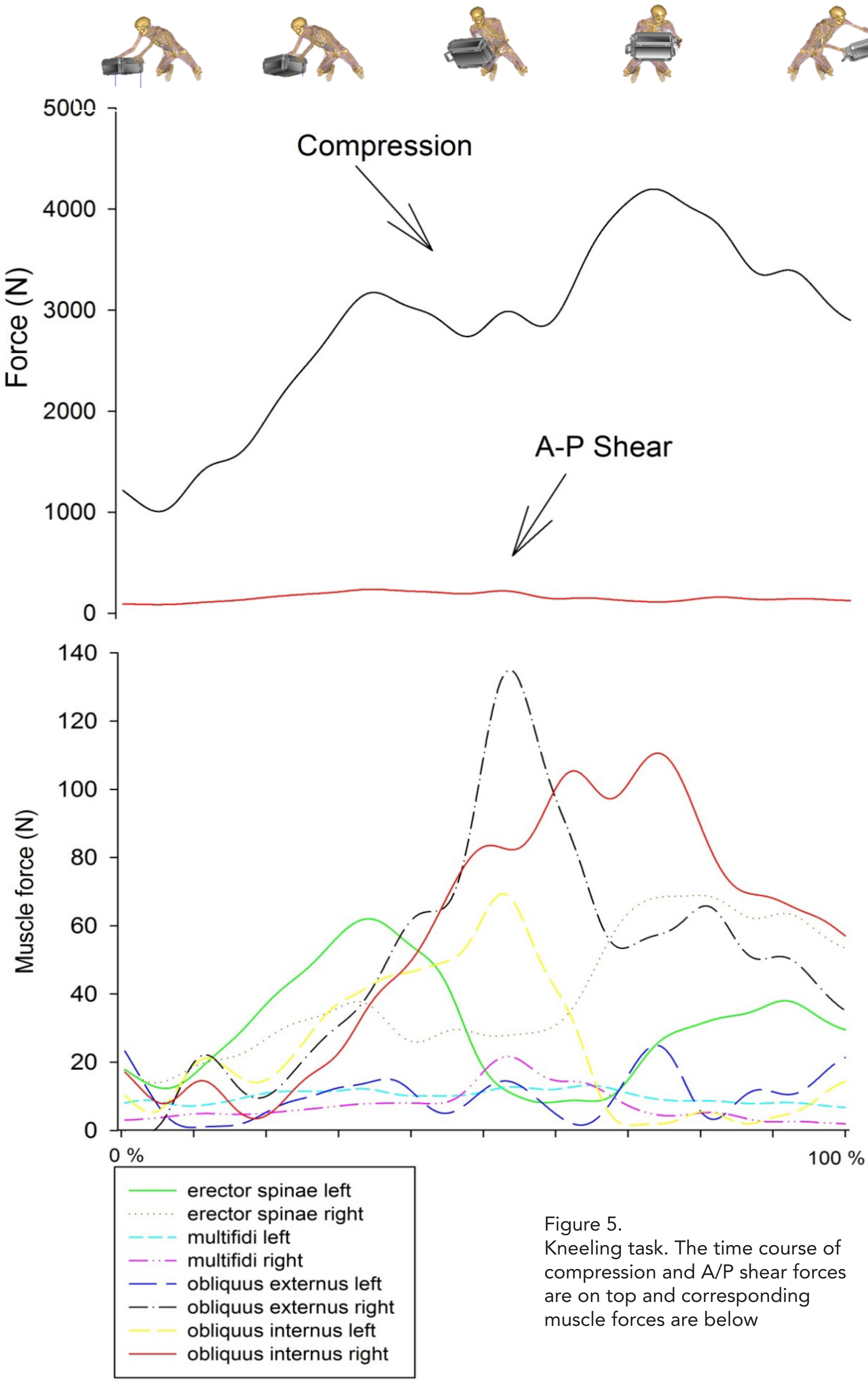

Figure 5.
Kneeling task. The time course of compression and A/P shear forces are on top and corresponding muscle forces are below

Figure 6 depicts the L4/L5 internal/external rotator moment and the force in the bilateral obliquus interna et externa. In the stooped position, the internal/external rotator moment had a high initial level from which the rotator moment decreased throughout the task (Figure 6). The initial large rotator moment was followed by a high force in all abdominal obliquus muscles except for the left obliquus externus. Similar to the internal/external rotator moment, the force in the obliquus muscles also declined throughout the task.
In the kneeling position, the first 2/3 of the task was dominated by a positive rotator moment (counter clock-wise) whereas the last part was dominated by a clock-wise rotator moment. When the moment increased, the force in the right obliquus externus and left obliquus internus increased accordingly. As the internal/external rotator moment turned negative (clock-wise dominance), the force in the right obliquus externus and left obliquus internus declined. The force in the left obliquus externus peaked simultaneously with the negative rotator moment peak (Figure 6).

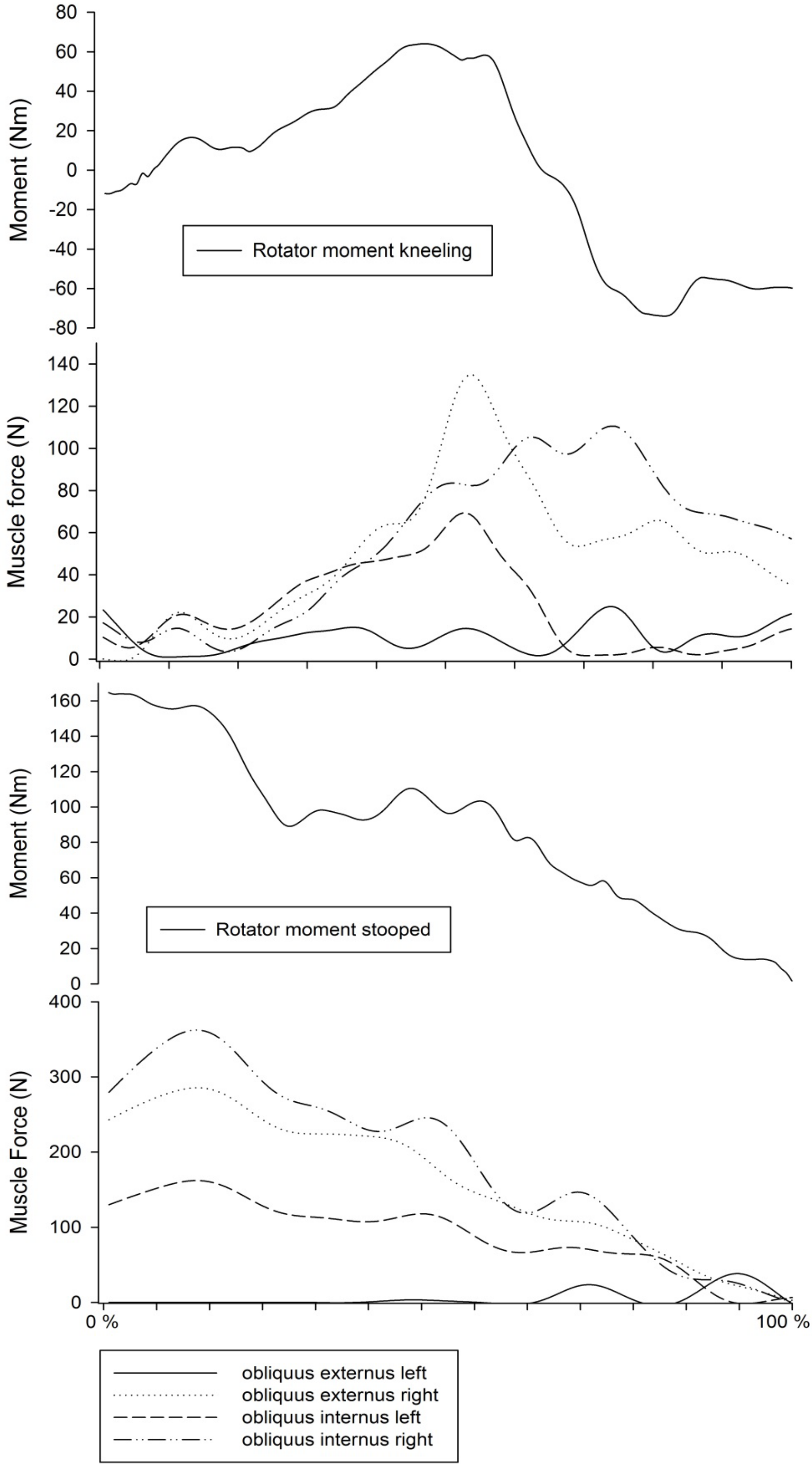

Figure 6. Internal/external rotator moment and muscle activity for kneeling and stooped tasks. Positive values indicate moments towards the subject's left

4.0 Discussion

This study investigated the spinal loading during heavy asymmetrical lifting using a highly detailed musculoskeletal computer model and found that the L4/L5 joint was compressed with up to 5541 N in a stooped lifting task and up to 4197 N in a kneeling lifting task. The A-P shear force peaked at 389 N in the stooped task and at 237 N in the kneeling task. The median compression value and IQR showed that the stooped task had a higher level of compression throughout the task. Hence, it may be advantageous to choose the kneeling position over the stooped position when possible.

Interestingly, the largest compression force was found in the stooped task with the lightest (10 kg) suitcase. This can probably be explained by the lifting technique. The subject performed the lift faster and accelerated the suitcase more, which increased the loading on the spine. Experimentally, this may have been avoided if the subject had been asked to perform the lifts at the same velocity despite the weight of the suitcase. However, this was considered an unnecessary and unrealistic intervention. We found that the median compression force was largest in the 20 kg task, second largest in the 15 kg task, and smallest in the 10 kg task, implying that the general loading in a task was larger with a heavier suitcase.

We found an L4/L5 internal/external rotation moment of more than 160 Nm in the stooped task (Figure 5), which was primarily created by mm. obliqii. Besides an internal/external rotator moment, these muscles also create a flexor moment in L4/L5. This flexor moment must be matched by an equivalent extensor moment created by m. erector spinae to maintain the standing position. This extensor moment also contributed to the compression in the L4/L5 joint.

Jäger et al. (13) found in their review of the literature that the estimated in vitro average compression tolerance for lumbar segments was 6180 N (SD 2660 N). Based on the compression forces from the model in the present study and the large variation of the estimate from Jäger et al. (13), compression injuries in the L4/L5 vertebrae are not unlikely to occur in baggage handling work. However, in vitro tolerance results may not be applicable for in vivo conditions. Some evidence exist that compression injuries to endplates and the underlying trabecular bone may be quite common and could be a cause for low back pain (25;26). Especially large compression forces cause these injuries (27) and repeated loading increases the risk of compression injuries (8;9;27). In an in vitro study, Brinckmann et al. (8) found a 55 % risk of sustaining a compression injury if a segment was repeatedly loaded 500 times with 40-50 % of the maximum compression tolerance. This could possibly explain the high prevalence of low back pain in different occupational groups with frequent heavy lifting (19;28;29).

Previously, compression forces of up to 36000 N in the lumbar spine have been reported in competitive weightlifters (30;31). These compression values were calculated with static 2D models and therefore underestimated. The reason why weightlifters do not develop compression fractures may be that they are highly selected with large discus area and may have developed high bone mineral density (31). The bone mineral content in the lumbar segments has a large influence on the compression tolerance. A cadaver study has shown that the compression tolerance increased with 1685 N when the bone mineral content increased by one g/cm3 (12). Moreover, the area of the discus has great influence on the intra-discal pressure. If the discus area is large, the compression force will be distributed on a larger area hence reducing the pressure on the discus. However, even if weightlifters have large discus areas and high bone mineral density, no spinal segments have been shown to withstand 36000 N compression in in vitro measurements. Therefore, additional mechanisms in the living organism must relieve the compression of the lumbar spine. The IAP may play an important role in reduction of the compression forces in the lumbar spine. It has been suggested that the IAP can reduce the compression force with a passive extensor moment (32;33). However, this will not be detectable in spinal models, which do not include a specific IAP model, as only net moments are accounted for in inverse dynamic analysis. Another possibility may be that the IAP acts as a semi-rigid cylinder on which the load from the upper extremities and thorax can rest (32). This will enlarge the target area for the compression from the area of the disc to the cross-sectional area of the trunk and therefore reduce the pressure on the spine markedly (32).

The shear forces found in the present study must be considered rather modest. In a review of the literature, Gallagher & Marras (34) found that appropriate limits for shear forces were 1000 N for few (<100) cycles per day and 700 N for frequent shear loading based on in vitro measurements of shear strength with no regard to other factors. Compared to these limits, a risk of developing injuries due to shear loading is not present. However, the results from the present study may be highly dependent on the definition of the orientation of the L5 coordinate system, therefore the sensitivity of shear forces to changes in the orientation of the L5 coordinate system should be investigated in more detail. We used the current orientation, because it was validated for compression forces (15). However, there is no report on the validity of the shear forces in the present model.

Many risk assessments of manual material handling are based on the NIOSH recommendations. However, the drawbacks of these recommendations cannot be disregarded. The NIOSH recommendations are partly based on static 2D computations of lifting (11). This is a coarse simplification and these computations are considered less valid than dynamic 3D models. The NIOSH recommendations do not account for sex and age although evidence suggests that women have a lower compression tolerance than men and that the compression tolerance declines with age (12;13;18;25). The age and sex of the individual are taken into account in the German Dortmund recommendations (13). These recommendations are based solely on in vitro measurements of compression tolerance and do not rely on inadequate biomechanical assumptions. However, they do not account for mechanisms in the living human body that may relieve the compression on the spine. NIOSH recommends a limit of 3400 N for continuous work and 6400 N for single lifts (11). In the Dortmund recommendations, the limits of compression are divided into sex and age in decades (13;18). Hence, the subject in the present study (a 48-year-old male) would be allowed between 4100 N (at 40 years) and 3200 N (at 50 years) compression during continuous work. However, since he is closer to 50 than 40 years, the individual recommendation should be around 3400 N just like the NIOSH recommendation. The compression forces found in the present study exceeded the recommended limits in both the Dortmund and NIOSH recommendations.

In conclusion, we estimated joint- and muscle forces during dynamic, asymmetrical lifting tasks using a musculoskeletal computer model. The present model is not limited to the two tasks in the present study but can be applied to any type of lift. The compression forces from the present musculoskeletal model exceeded the recommended limits except for the kneeling lift of 10kg and 15 kg suitcases. Although average compression tolerance reported in the literature was not exceeded, compression injuries in the L4/L5 vertebra are not unlikely to occur at the compression forces found in this study.

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