

介護作業における腰への負荷の動的3D生体力学評価

A dynamic 3D biomechanical evaluation of the load on the low back during different patient-handling tasks

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抄録（仮訳）

この研究の目的は、患者の介助に係る一般的な作業について、腰への負荷を調査することである。患者の介助について正式なトレーニングを受けていない10人の女性医療従事者が、男性の脳卒中患者に対し回転、持ち上げ、位置修正を含む9つの介助作業を実施した。このときの腰への負荷を、脊柱起立筋の筋電図（EMG）と自覚的運動強度（RPE：Borgスケール）により測定し、下位腰椎（第4-第5腰椎椎間関節）に作用する有効モーメント、圧縮・せん断力を定量化した。実験は5台のカメラを用いたビデオ・システムでビデオ録画（50コマ/秒）され、医療従事者の床とベッドサイドへの反作用力は、床反力測定盤とベッドに取り付けた力変換器によって記録された。

生体力学的負荷は、下半身の動的な3D-7分割モデルを使って計算され、第4-第5腰椎椎間関節に作用する力は腰部の14の筋肉の断面モデルで推定された。（最適化手順）

圧縮力とトルク(ねじれモーメント)は、介助作業の内容から強く影響を受けるのに対し、**筋電図と自覚的運動強度は、介助を行う被験者に依存**していた。患者

(4132/4433N)の持ち上げを含む2つの作業中の最大圧縮力は、他の全ての作業中より有意に高かった。ベッド(3179/3091/2932/3094N)での患者の位置修正を含む4つの作業に差はなかったが、2つのベッド(1618/2197N)での患者の回転作業に比べて高い最大圧縮力を示した。

このように、本研究において、患者の取扱い作業は、第4-第5腰椎椎間関節に瞬間的に作用するねじれモーメントと圧縮力の違いにより、持ち上げ、位置修正、回転の3つのグループに分類された。

A dynamic 3D biomechanical evaluation of the load on the low back during different patient-handling tasks

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Abstract

The objective of this study was to investigate the low-back loading during common patient-handling tasks. Ten female health care workers without formal training in patient handling performed nine patient-handling tasks including turning, lifting and repositioning a male stroke patient. The low-back loading was quantified by net moment, compression, and shear forces at the L4/L5 joint, measured muscle activity (EMG) in erector spinae muscles and rate of perceived exertion (RPE; Borg scale). The experiments were videotaped with a 50 Hz video system using five cameras, and the ground and bedside reaction forces of the health care worker were recorded by means of force platforms and force transducers on the bed. The biomechanical load was calculated using a dynamic 3D seven-segment model of the lower part of the body, and the forces at the L4/L5 joint were estimated by a 14 muscles cross-sectional model of the low back (optimisation procedure). Compression force and torque showed high task dependency whereas the EMG data and the RPE values were more dependent on the subject. The peak compression during two tasks involving lifting the patient (4132/4433 N) was significantly higher than all other tasks. Four tasks involving repositioning the patient in the bed (3179/3091/2932/3094 N) did not differ, but showed higher peak compression than two tasks turning the patient in the bed (1618/2197 N). Thus, in this study the patient-handling tasks could be classified into three groups—characterised by lifting, repositioning or turning—with different levels of peak net torque and compression at the L4/L5 joint. © 2002 Elsevier Science Ltd. All rights reserved.

Keywords: Patient handling; Health care worker; Low back; Biomechanical load

1. Introduction

A high prevalence rate of low-back disorders (LBD) is found among health care workers (HCW) (Harber et al., 1985; Jensen et al., 1995; Jensen and Tüchsen, 1995; Pheasant and Stubbs, 1992; Stubbs et al., 1983). Compared to other jobs with heavy lifting tasks and a high prevalence rate of LBD, e.g. construction workers, the total weight lifted per day is lower in the health care sector. This may indicate that the risk in connection with a single personal handling situation is higher than the risk in connection with lifting, e.g. a box. Results from epidemiological studies show strong evidence for an association between manual material handling, frequent bending and twisting, physically heavy work, whole body vibration and LBD (Bernard, 1997; Hoogendoorn et al., 1999). Several of these risk factors

are present in patient-handling tasks. Biomechanical studies have estimated the load on the low back in several patient-handling tasks (Daynard et al., 2001; de Looze et al., 1994; Dehlin and Lindberg, 1975; Gagnon et al., 1986, 1987, 1988; Garg et al., 1991a, b; Garg and Owen, 1993; Lindbeck and Engkvist, 1993; Ulin et al., 1997; Winkelmoen et al., 1994). However, no detailed knowledge exists concerning the level of exposure at which the risk for LBD increases. To get a more detailed knowledge concerning the risk factors in the health care sector, exposure assessment may be based on a description of the type and amount of tasks each worker performs in future epidemiological studies. If the internal mechanical load of each type of tasks being performed is known, the internal dose can be calculated. Measurement of the internal mechanical load during different tasks is very time consuming and can only be performed on a small number of subjects. The precondition for a valid exposure description based on this method is a low variance in the mechanical load between

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subjects performing the same task compared to the difference between tasks. Psychophysical exposures in epidemiological studies basically face the same problems; however, psychophysical measurements are easier to perform on large subject groups. Accordingly it is relevant to investigate a possible correlation between psychophysical measures for the low back and the mechanical load during patient-handling tasks. Hence, the objectives of this study were (1) to estimate the mechanical load on the low back during common patient-handling tasks; and (2) to compare the variation in the load between the different tasks with the variation in load between subjects when they perform the same task; and finally (3) to monitor the relations between the mechanical load and psychophysical measures for the low back during patient-handling tasks.

Most studies on the low-back loading of HCWs during patient-handling tasks have been done for symmetrical loading conditions with 2D models and with a healthy subject or manikin assuming the role of a patient (de Looze et al., 1994, 1998; Gagnon et al., 1986, 1988). In this study nine patient-handling tasks were evaluated with a disabled (stroke) person. The spinal loading was calculated by means of a dynamic 3D biomechanical model of the lower part of the body and measurement of reaction forces from the ground and from the bed of the patient (Skotte, 2001). The joint forces of the L4/L5 joint were estimated by a 14 muscles cross-sectional model of the low back, and by minimising the sum of cubed muscle stresses.

2. Methods

2.1. Subjects and tasks

The study included 10 female HCWs (Table 1). Their job experience with patient handling was 19 (6–26) years, and in the present job they handled 7 (2–20) patients per day, but they have had no special education or training with respect to patient-handling technique. All HCWs gave informed written consent, and the study was approved by the local Ethical Committee. None of the HCWs experienced low-back pain on the experi-

mental day. In a laboratory set-up they performed nine different patient-handling tasks (Fig. 1) in randomised order. No instructions were given prior to performing the tasks, but the HCWs were asked to handle the patient with the techniques they used during normal work and to use a normal pace. No assistant devices were used, but in the tasks 2 and 9, some of the HCWs preferred to get assistance from a second HCW. The male patient had suffered from stroke, was 53 years old, 1.75 m high and had a body weight of 88 kg. He had spastic paralysed muscles in both sides primarily in the left side, but had a normal function of the right arm. He was instructed not to offer resistance to the HCW but only to co-operate, when he was instructed so by the HCW. The bed had a design that is typically found in a hospital in Denmark, and the height of the bed could be mechanically adjusted. In addition to the nine patient-handling tasks, the HCW carried out a standardised symmetrical reference lift (SL) by lifting a 15 kg load from the middle of the bed to elbow height near the body and back again. At the beginning of the lifting, the horizontal distance from the tip of the toes to the handles of the load was 40 cm, and the height of hands from the floor was 85 cm. This standardised lift was included in order to compare the load during the different patient-handling tasks with a simple lifting situation causing a low-back load at a level just acceptable by the Danish Working Environment Authority.

2.2. Measurements

The load on the low back during the patient-handling tasks was quantified by biomechanical calculation of torque, compression, and shear force at the L4/L5 joint, the measured muscle activity (EMG) in erector spinae muscles, and the rate of perceived exertion.

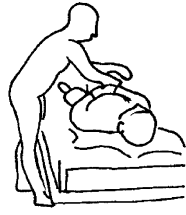
A dynamic 3D biomechanical model of the lower part of the body including feet, legs, thighs, and pelvis was used for calculating the net torque at the L4/L5 joint. Ground reaction forces were measured in three directions by means of two force platforms with the HCW standing with one leg on each platform or both legs on the same force platform. In addition, to measure the horizontal reaction force from the bed, when the HCW exerted a pressure on the bed with her leg, the bedside was fitted with two force transducers connected by a bar. The experiments were videotaped with a 50 Hz video system using five cameras, and digitised automatically with a Peak Motus 4.3 system. Details on marker positions and the biomechanical model for calculating the torque at L4/L5 can be found in Skotte (2001). The muscle model and the method used for estimating compression and shear forces are described in the appendix. The biomechanical analysis was carried out for the central part of the tasks while the HCW

Table 1
Subject data

<i>N</i> = 10	Mean	SD	Minimum	Maximum
Age (years)	43	8.7	33	59
Height (cm)	169	5.8	161	178
Weight (kg)	72	12.1	54	95
Pelvis width (cm) ^a	24.7	1.8	21	27
Pelvis depth (cm) ^b	17.8	1.5	15	20

^aDistance between right and left anterior superior iliac spine.

^bDistance between anterior and posterior superior iliac spine.



1. Turning the patient in bed towards the HCW from his back to his left side (causing his face to turn towards the HCW).



6. Moving the patient from sitting on the bed to supine position in the bed.



2. Repositioning of the patient from lying in the middle of the bed to the bedside nearest the HCW.



7. Transferring the patient from sitting on the bed to sitting in a wheelchair.



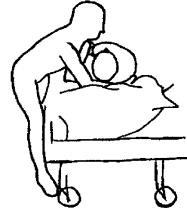
3. Turning the patient in bed away from the HCW from his back to his right side.



8. Repositioning the patient posteriorly in the seat of the wheelchair.



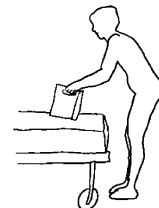
4. Elevate the patient from a supine position in the bed to sitting position on the edge of the bed.



9. Repositioning of the supine patient towards the head of the bed.



5. Lifting the patient from sitting on the edge of the bed to standing on the floor.



SL. Standard lift.

Fig. 1. The patient handling tasks.

moved the trunk of the patient. It was considered that the maximum exertion of the HCW would occur in this part of the process, which ranged from 2 to 17 s. It was not feasible to carry out the biomechanical analysis of the preparatory doings like adjusting the bed, positioning the arms of the patient, etc. Peak value of torque,

compression and shear force at L4/L5 during the central part of each task were chosen as descriptive variables. The coordinate system was oriented with the x -axis anteriorly, the y -axis left laterally and the z -axis superiorly relative to the disc centroid. No biomechanical calculations were performed for task 7 due to technical

problems in measuring the reaction force with the force plates during movement of the patient from the bed to the wheelchair.

Pre-gelled Ag/AgCl surface electrodes were used to measure the EMG activity of the erector spinae muscles during the patient-handling procedures. They were placed 3 cm apart on both sides of L3, approximately 3 cm lateral to the spinal column in the middle of the muscle bulb. Before carrying out the tasks, EMG was measured when the HCWs performed three maximal isometric contractions (MVC) for the back extensor muscles standing in the upright position. The highest obtained EMG value during these contractions, EMG_{ref} , was used to normalise the EMG registrations during the tasks. There are considerable limitations in this method of normalisation; however, this was the available possibility in order to get a rough estimate of the muscle activation. It was not the aim to estimate muscle forces from the EMG data. The EMG signals were pre-amplified, low-pass filtered at 450 Hz and sampled with a frequency of 1000 Hz. The recorded data was high-pass filtered with a cut-off frequency of 10 Hz, rectified and low-pass filtered with a second-order single-pass Butterworth filter with a cut-off frequency at 2.5 Hz (Brereton and McGill, 1998). The peak EMG and the time period when EMG was above 50% EMG_{ref} were chosen as descriptive variables.

Immediately after completion of each patient-handling task, the HCW was asked to rate her perceived physical exertion (RPE) on the low back by answering the question 'how did you perceive the exertion on the low back'. The Borg CR10 scale (Borg, 1990), where 0

implies 'nothing at all' and 10 implies 'extremely strong' perceptual intensity, was used for the rating.

2.3. Statistics

After testing the residuals for normal distribution and variance homogeneity, a two-way ANOVA was used with tasks and HCWs as fixed factors. The significance level was set at $p < 0.05$. To compare the variance between tasks with the variance between HCWs, the sum of square of variance (SS) for tasks and for HCWs were expressed as a percentage of the total SS. When significant differences were found and the variance due to tasks exceeded the variance due to HCWs, a multiple comparison test (Tukey Test) was performed to isolate which task(s) differed from the others. Finally, a grouping of the tasks according to total net torque and compression force was analysed by means of Scheffé's test for multiple contrasts (Zar, 1996).

3. Results

Fig. 2 shows an example of the calculated net torque (three components) and compression force at the L4/L5 joint during a 4 s period of task 4. This example shows a patient-handling situation with an asymmetric loading causing a peak compression of 3330 N on the low back. Summary results of biomechanical calculations, EMG recordings, and RPE data are shown in Table 2. Fig. 3 shows individual peak compression values for all HCWs and tasks. Individual peak net torque and compression

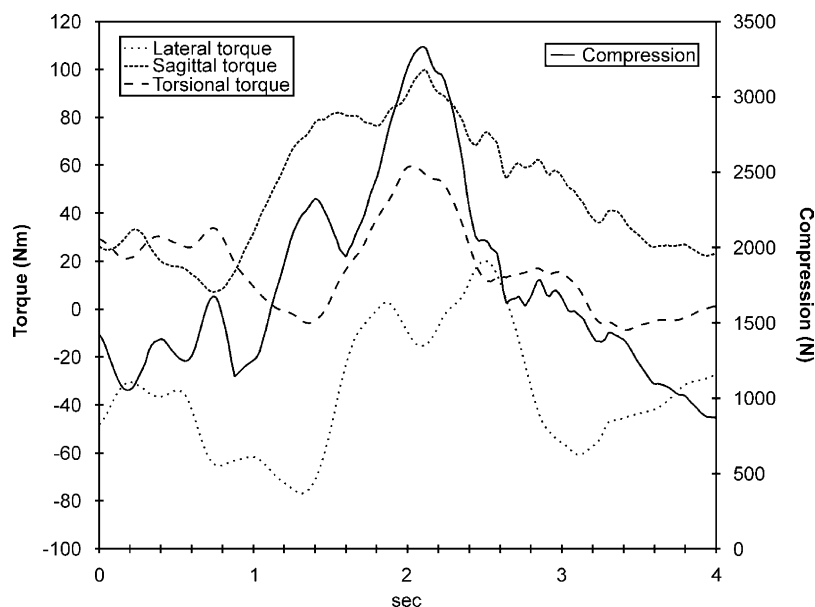


Fig. 2. Net torque (left vertical axis) and compression (right vertical axis) at the L4/L5 joint of HCW 5 for task 4 (elevating the patient from a supine position in the bed to a sitting position on the edge of the bed). For the sake of convenience compression values are shown as positive.

Table 2
Mean, standard deviation and range of low back (L4/L5) torques, forces, EMG recordings, and RPE values

Tasks	1	2	3	4	5	6	7	8	9	SL ^a
Total torque (Nm)	73±20 ^b (52–105)	148±38 (77–225)	97±33 ^b (65–170)	136±29 (107–187)	192±33 ^c (149–259)	125±26 (95–176)	N.A.	194±28 ^c (143–239)	143±37 (95–204)	146±24 (105–189)
Lateral torque (Nm)	19±5 (11–29)	33±9 (20–48)	30±15 (10–65)	64±13 ^c (44–83)	40±25 (16–107)	40±12 (27–64)	N.A.	41±21 (17–72)	32±9 (21–49)	12±7 (5–28)
Extension torque (Nm)	71±20 ^b (51–103)	146±38 (73–222)	95±33 (59–167)	125±32 (89–183)	184±42 ^c (106–259)	116±33 (69–174)	N.A.	189±31 ^c (138–238)	139±40 (85–204)	145±24 (104–189)
Torsion torque (Nm)	15±7 (7–25)	29±12 (15–49)	19±7 (11–36)	48±12 ^c [6] (34–70)	23±14 (8–55)	37±13 (17–54)	N.A.	18±7 (3–26)	25±12 (9–42)	8±4 (4–16)
Ant/pos shear force (N)	302±67 (209–413)	425±106 (282–661)	258±76 (129–334)	405±78 (259–502)	252±75 (106–352)	352±82 (222–490)	N.A.	314±116 (142–576)	353±121 (136–588)	238±54 (140–306)
Lateral shear force (N)	63±21 (39–102)	141±44 (52–210)	115±75 (58–317)	192±58 ^c [2] (113–277)	127±46 (71–209)	210±67 ^c (99–294)	N.A.	116±42 (51–208)	123±39 (51–192)	55±11 (40–78)
Compression (N)	1618±322 ^b (1139–2114)	3179±631 (2256–4427)	2197±585 ^b (1396–3455)	3091±412 (2470–3914)	4132±631 ^c (3313–5162)	2932±450 (2165–3757)	N.A.	4433±666 ^c (3464–5509)	3094±591 (2316–3982)	3099±402 (2388–3772)
Peak RESEMG ^d (%)	77±29 (29–123)	132±72 (52–293)	124±76 (34–277)	129±76 (55–252)	130±52 (34–212)	146±51 (77–231)	118±40 (58–201)	129±65 (62–277)	210±79 (78–297)	111±39 (43–176)
Peak LESEMG ^e (%)	78±54 (41–220)	118±76 (40–302)	113±59 (29–200)	118±79 (80–335)	154±65 (74–260)	100±35 (61–144)	141±78 (66–329)	137±57 (58–239)	147±71 (73–305)	100±32 (55–148)
RESEMG > 50% ^d (s)	1.8 (0–5.8)	3.4 (0.1–9.4)	3.0 (0–9.7)	4.7 (0.2–10.4)	2.7 (0–6.7)	3.9 (0.7–9.5)	6.6 (0.4–27.1)	2.5 (0.3–6.1)	4.6 (1–16.4)	1.5 (0–3.5)
LESEMG > 50% ^e (s)	1.3 (0–7.5)	2.7 (0–5.0)	2.5 (0–7.9)	4.8 (2–11.5)	2.4 (1.2–5.1)	2.6 (0.3–8.1)	4.1 (0.9–9.5)	2.1 (0.5–3.9)	3.1 (1.1–7.9)	1.4 (0.1–3.3)
RPE (Borg CR10 scale)	1.3 (0–4)	2.1 (0–4)	1.8 (0–6)	2.6 (0–7)	1.9 (0–7)	2.8 (0–8)	2.1 (0–7)	1.8 (0–5)	2 (0–7)	1 (0–3)

^aSL—Standard symmetrical lift.
^bSignificantly smaller than other tasks.
^cSignificantly larger than other tasks except tasks mentioned in [].
^dRESEMG—right erector spinae EMG.
^eLESEMG—left erector spinae EMG.

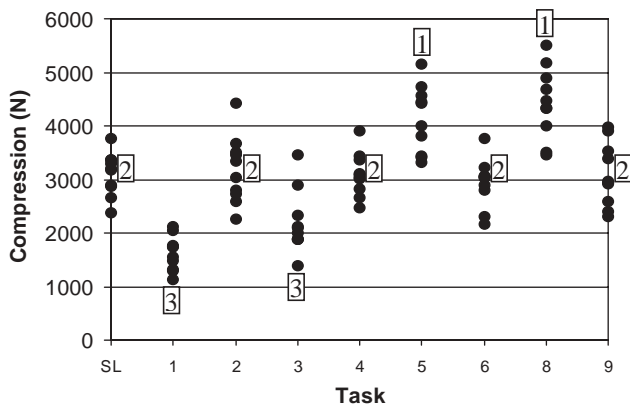


Fig. 3. Peak compression values for all HCWs and tasks (SL—standard symmetrical lift). The numerals 1, 2, and 3 in the quadrangles indicate a classification of the tasks on the basis of the compression values (see the discussion).

values for all trials ranged from 52 to 259 Nm and from 1283 to 5509 N, respectively. The highest mean of the peak compression values was found for task 5 (4132 N) and task 8 (4433 N), and they did not differ significantly. Also the task mean compression values for tasks 2 (3179 N), 4 (3091 N), 6 (2932 N), and 9 (3094 N) did not differ but were significantly lower than for tasks 5 and 8. The lowest task mean compression values was found

for task 1 (1618 N) and task 3 (2197 N). Peak anterior/posterior and lateral shear forces were found in the range 106–661 and 40–317 N, respectively. Except for one trial all peak anterior/posterior shear forces were pointing forward. The total range of measured peak EMG values was 29–335% relative to EMG_{ref} and the time period when EMG exceeded 50% of EMG_{ref} ranged from 0 to 27.1 s. The range of RPE values was 0–8.

The distribution of the variance from the HCW and the task is shown in Table 3. For every variable measured, the difference in the mean values among the tasks was greater than would be expected by chance after allowing for effects of differences in HCWs ($p \leq 0.01$). However, in almost every measure (except lateral and torsional torque ($p > 0.1$)), the difference in the mean values among the HCWs was also greater than expected by chance after allowing for effects of differences in tasks ($p < 0.004$). The EMG data were found to be more dependent on the HCW than the biomechanically calculated parameters. No correlation was found between the EMG and the RPE values, or between the compression forces and the RPE values. In the tasks 1, 2, 3, 4, 6, and 9, the two highest compression values were found for HCW 3 and 8. These two HCWs also had the highest bodyweight (85 and 95 kg, respectively) of all the HCWs.

Table 3
The sum of squares of variance (SS) for the tasks and the subjects as percentage of the total SS

	% variance due to task	% variance due to subject
Subject		
Total torque	63.9	20.0
Lateral torque	42.1	3.5
Extension torque	57.9	22.1
Torsion torque	50.4	9.5
Ant/pos shear force	31.3	40.7
Lateral shear force	43.9	17.4
Compression	73.1	8.7
Peak RESEMG ^a	24.1	44.7
Peak LESEMG ^b	20.0	49.8
RESEMG > 50% ^a	10.2	56.4
LESEMG > 50% ^b	15.7	61.3
RPE	4.7	83.0

^a RESEMG—right erector spinae EMG.

^b LESEMG—left erector spinae EMG.

4. Discussion

4.1. Biomechanical load

The analysis of variance indicated that the biomechanically calculated parameters are more dependent on the task than on the HCW. This was most pronounced for the low-back compression and the total torque. If the subjects in this study are representative for the HCW staff in Denmark, these findings are remarkable since this gives the possibility of making exposure descriptions of torque and compression, by describing the tasks each HCW perform. As illustrated in Fig. 3, it is reasonable to group the tasks into three classes:

- Class 1, high compression (mean 4283 N): tasks 5 and 8.
- Class 2, medium compression (mean 3074 N): tasks 2, 4, 6, and 9.
- Class 3, low compression (mean 1907 N): tasks 1 and 3.

The mean of the peak compression values is significantly higher for class 1 than for class 2, and also for class 2 compared to class 3. Classification of the tasks by means of the total torque resulted in the same grouping of the tasks. The high low-back loading during the tasks in class 1 is found during a (vertical) lift of the patient. The medium load during the tasks in class 2 occurs during horizontal repositioning of the patient in the bed or during repositioning the patient from supine to sitting in the bed and vice versa. The low load on the low back during the tasks in class 3 is found when the patient is turned in bed. For practical reasons no biomechanical data were acquired for task 7. However, by using the above results this task should be assessed as

a class 1 task because the task includes a lift of the patient from sitting at the bedside in the same way as lifting the patient from sitting to standing in the tasks 5 and 8. The weight of the HCW exerted a noticeable influence on the low-back compression as the two HCW with the highest weight (HCW 3 and 8) also showed the highest compression in all of the tasks in class 2 and 3 but not in class 1. It is not surprising that the compression is influenced by the weight of the trunk when the external load is relatively small. Though the biomechanical parameters showed a pronounced task dependency, it should be noticed that techniques and the use of assistance devices differ in the patient-handling situations in the health care sector and it can be difficult to compare different techniques used to carry out a task. As an example, task 9 was carried out in the following different ways: 5 HCWs used a two-person hook method, 3 HCWs used the one-person hook method and 2 HCWs used other one-person methods standing at the head or foot of the bed.

The compression data are consistent with or somewhat lower than the data from the most of earlier studies using similar methods. The choice of a real patient in this study instead of using a manikin or a healthy person as patient as in many earlier studies (de Looze et al., 1994, 1998; Gagnon et al., 1986, 1988) could to some extent account for these differences. This choice was made in order to make the cooperation between HCW and patient realistic. Furthermore it would make the HCW base the whole handling situation on the actual physical resources of the patient. Earlier findings confirm that patient-handling tasks including (vertical) lifts, result in compressions between 4000 and 5000 N (de Looze et al., 1994; Gagnon et al., 1986; Garg et al., 1991a, b; Garg and Owen, 1994). Patient-handling tasks including horizontal repositioning in the bed result in compressions around 4000 N, however, only represented by 'repositioning a patient towards the head of the bed' (de Looze et al., 1994). Patient-handling tasks including turning the patient in bed result in compressions between 2500 and 3500 N (de Looze et al., 1994; Gagnon et al., 1987). Recently Daynard et al. (2001) reported spinal peak loadings similar to the results in this study: For untrained HCWs the peak compression for tasks designated as 'bed turn', 'chair boost' and 'bed boost' were 2129, 4766 and 3283 N, respectively, compared to 1618/2197, 4433 and 3094 N for tasks 1/3, 8 and 9 in this study. In task 5, lifting the patient from sitting on the bed to standing on the floor, Gagnon et al. (1986), de Looze et al. (1994) and the present study found peak compression values very close to 4000 N. The reason why the results in particular agree so well in task 5 is probably due to a very symmetric lifting situation. The mentioned studies have all used a 2D model, but tasks with symmetric work situations should not deviate from the present results based on a

3D-model. Compression values much higher than the above values have been reported by Marras et al. (1999) who used an EMG driven model. The calculations in the Marras study showed compression forces in the range 6000–9000 N at L5/S1 during repositioning a supine person (weighing 50 kg) towards the head of the bed. It is difficult to explain this high discrepancy, although the differences could partly be caused by muscle coactivity, which the method used in this study, did not account for. de Looze et al. (1998) found peak torque values of 212 ± 56 N m during lifting the patient from sitting on the bed to standing on the floor, which are in good agreement with the present results of 192 ± 33 N m (task 5). However, de Looze did not find any differences in peak torque during turning the patient (197 ± 57 N m) in bed compared to lifting the patient from sitting on the bed to standing on the floor. In this study the peak torque during turning (task 1: 73 ± 20 N m, task 3: 97 ± 33 N m) was less than half the value during lifting the patient (192 ± 33 N m). A reason for this discrepancy between the studies could possibly be found in different ways of measuring the reaction forces on the HCW during the performance of the tasks. Both studies used a biomechanical model of the lower part of the body and measured ground reaction forces by force platforms. Furthermore, in the present study, the reaction forces between the bedside and the HCW were measured and included in the calculation of the torque. Nearly all of the HCWs exerted a force on the bedside during turning the patient in bed (tasks 1 and 3). Failing to include this bedside reaction force in the calculation could lead to a substantial over-estimation of the torque (Skotte, 2001). Very large shear forces have been reported during patient handling (Gagnon et al., 1986, 1988; Marras et al., 1999), 2–3 times larger than in the present study and what was found by de Looze et al. (1994). The calculated shear forces are very dependent upon the anatomical model chosen (Parnianpour et al., 1997), and these discrepancies may therefore to a large extent be due to differences in the biomechanical models.

This study utilised a 3D dynamic model of the lower part of the body that made it possible to include asymmetric loading and rapid movements in the biomechanical calculations. Two tasks caused more asymmetrical loading on the low back than the rest of the tasks. Task 4 had a significantly higher lateral torque than all the other tasks. The amount of asymmetrical loading calculated as the root mean square of lateral and torsional torque relative to extension torque was higher for task 4 (64%) and task 6 (47%) than for the other tasks (24–37%). These results are in good agreement with the observed posture of the HCWs during the patient handling. In the example in Fig. 2, which shows a situation with asymmetric loading, the torsional torque has a significant influence on the compression. The peak compression is calculated to 3330 N, however,

it would be 2290 N if the calculation was made without the inclusion of the lateral and torsional components of the torque.

Studies on the low-back loading of HCWs often have been made with static models. In this study the dynamic calculations could be compared to quasi-static calculations not including linear and angular acceleration of the lower part of the body. In one case the difference between dynamic and quasi-static calculation of the torque was 17% for a situation where the trunk of the patient was repositioned with a sudden pull. However, in 90% of the cases, the difference was less or much less than 5%. These small differences were expected since the lower part of the body normally is fairly stationary compared to the upper part of the body during patient-handling tasks. Moreover, most tasks are performed slowly.

According to the revised NIOSH equation (Waters et al., 1993), patient-handling tasks included in the classes 2 and 3 should be safe to carry out for most workers, as the proposed safety limit of 3400 N should protect 99% of male workers and 75% of female workers. However, only in task 1 the maximum peak compression force did not exceed this limit. Other authors suggest taking into consideration that age and gender are factors that influence a person's physical capacity (including spinal strength). Jäger and Luttmann (1997) therefore proposed age and gender specific limits based on biomechanical findings. As an example, they proposed limits for women ranging from 4400 N at the age of 20, 3200 N at the age of 40 and finally to 1800 N at the age of 60 or more. According to these limits, tasks including (vertical) lifting (class 1) as well as tasks including horizontal repositioning of the patient (class 2) may be hazardous to women from their forties who are highly represented as well in the present study as in the health care sector.

4.2. EMG

The EMG data were found to be more HCW dependent than the biomechanically calculated parameters. This is not surprising since the EMG data reflects individual differences in muscle activation, which the optimisation procedure does not account for. However, also the normalisation method used in this study could play a role because the length of the muscle and the velocity of the shortening effect the EMG–force relationship, and our normalisation does not take this into account. The EMG recordings could be more adequate utilised if an EMG driven model for estimation of the load had been available. Only one study was found in the literature on patient handling showing EMG peak values with reference to static MVC. In three different ways of turning the patient in the bed, Gagnon et al. (1998) found that erector spinae

maximally performed 66–74% of EMG_{ref} . This is in accordance with task 1, but somewhat lower than the values in task 3 in the present study. However, this is not surprising since the range of the results is large in accordance to the HCW dependence, and only 6 HCWs participated in the study by Gagnon.

4.3. Rate of perceived exertion

The RPE was evaluated by means of Borg's category-ratio (CR) 10 scale, which is designed for rating exertion in isolated body regions (Borg, 1990). The RPE values on the low back, which were much more dependent on the HCW than on the tasks, were generally low, but single individuals did perceive noticeable physical exertion. The great variance between HCWs may partly be explained by differences in performing a task. No correlation was found between the EMG and the RPE values, or between the compression forces and the RPE values. As an example, the two tasks with the highest RPE score (tasks 4 and 6) were tasks with intermediate compression values (class 2 tasks). Although this study could not show a correlation between the RPE scores and the estimated compression force of the low back, psychophysical measures could be a valuable supplement to the biomechanical measures.

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Appendix A

To estimate the compression and the shear forces at the L4/L5 joint, a cross-sectional model of the low back with 14 muscles was applied: right and left erector spinae, latissimus dorsi, internal obliques, external obliques, rectus abdominis, quadratus lumborum, and psoas. Muscle cross-sectional areas and moment arms were obtained from Chaffin et al. (1990) who provided data on lumbar muscle sizes and locations from CT-scans of females with an age approximately matching the HCWs in this study. However, according to McGill et al. (1996) the moment arms of erector spinae and rectus abdominis were increased with 13% and 30%, respectively, to account for the fact that CT-scans are carried out with the subject in a supine position. The muscles were modelled as single force vectors and the lines of action were derived from Dumas et al. (1991) who provided data on the orientation of several muscles for the trunk in a normal upright posture. In this study the orientation of the muscles were considered to be at a

constant angle with the plane of L4/L5, and representing a posture with a 15–20° flexion of the low back. Data on anatomical muscle cross-sections, orientation, and moment arms used in this study are shown in Table 4. The mean moment arms in Table 4 were scaled individually to the measured pelvis depth and width of the subject. Physiological cross-sections were calculated from the anatomical cross-sections by multiplying with the absolute value of the longitudinal component of the unit force vector. In order to derive compression and shear forces from net inter-segmental reaction torques, a numerical optimisation method was used. The magnitude of muscle forces f_i ($i = 1, 2, \dots, 14$) is constrained by $0 \leq f_i \leq d_i S_{max}$, where S_{max} is the maximum stress and d_i is the physiological cross-section of the muscles. The torque equilibrium conditions about L4/L5 are $\mathbf{A}\mathbf{f} = \mathbf{M}$ in matrix form, where \mathbf{M} is the reaction torque vector computed by means of a 3D dynamic biomechanical model. The vector \mathbf{f} consists of the muscle forces f_i , and \mathbf{A} is a 3×14 matrix with the columns $\mathbf{r}_i \times \mathbf{e}_i$, where \mathbf{r}_i is the moment arm vector and \mathbf{e}_i is the force unit vector for muscle i . The object function to be minimised is the sum of cubed muscle stresses $\sum (f_i/d_i)^3$, which according to Hughes et al. (1994) is able to predict muscles stresses in agreement with electromyographic data. The Matlab procedure *fmincon* was used to solve the optimisation problem. In the literature the maximum muscle stress S_{max} have been reported in the range 30–100 N/cm². In this study S_{max} was fixed to 90 N/cm² and the maximum uniaxial torque that the model could predict using the mean cross-sectional values was: flexion—105 N m, extension—208 N m, lateral bending—182 N m, and torsion—75 N m. In 9 of 91 trials, no feasible solution was found. However, by replacing the mean muscle cross-sectional values by mean plus 2 SD values, these 9 trials resulted in feasible solutions. A value of 90 N/cm² for S_{max} may be too high for the trunk musculature (McGill, 1991). However, by using a much lower value such as 35 N/cm², it was not possible to produce feasible solutions for most of the trials. The muscle model used in this study implies several simplifying assumptions: Muscle co-contractions and the resistance of passive tissues were not included and wide muscles such as the internal and external obliques, and latissimus dorsi muscles were represented as single-vectors. Moreover, the erector spinae musculature was modelled as one single muscle with a fixed orientation with regard to the orientation of the L4/L5 joint. In order to illustrate the effect of changing the value of the angle between the erector spinae muscle equivalent and the axis of the L4/L5 joint, this angle was increased with 15° (force unit vector changed to $[-0.34, 0.09, -0.93]$). This resulted in very small changes (mean value 1%) in compression but very large changes (mean value 67%) in anterior/posterior shear force. This shows that the estimated shear force is

Table 4

Data for anatomical cross sectional areas, moment arms, and line of action unit vectors for the right side of a low-back cross-section in the plane of L4/L5.

Muscle	ES	LD	IO	EO	RA	PS	QL
Anatomical cross-section (cm ²)							
Mean	17.4	1.4	5.3	5.3	4.1	9.8	4.6
SD	3.0	0.6	1.7	1.5	1.1	2.1	1.2
Moment arm (cm), sagittal plane	-6.0	-1.6	2.9	3.0	9.1	-0.2	-2.8
Moment arm (cm), frontal plane	-3.5	-11.9	-11.5	-12.2	-4.2	-4.4	-7.5
x-component of force unit vector	-0.09	-0.26	-0.71	0.17	0.0	0.41	0.17
y-component of force unit vector	0.09	0.37	0.0	0.34	0.03	-0.17	-0.50
z-component of force unit vector	-0.99	-0.88	-0.71	-0.92	-1.0	-0.89	-0.86

Orientation of coordinate system: The *x*-axis is directed anteriorly, the *y*-axis is directed left laterally, and the *z*-axis is directed superiorly relative to the disc centroid.

Notation of muscles: ES—erector spinae, LD—latissimus dorsi, IO—internal obliques, EO—external obliques, RA—rectus abdominis, PS—psoas, and QL—quadratus lumborum.

very dependent of the model used for the erector spinae musculature.

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