Lumbar-Load Analysis of Manual Patient-Handling Activities for Biomechanical Overload Prevention Among Healthcare Workers

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Received 25 April 2012; in final form 17 September 2012; Advance Access publication 18 December 2012

Manual patient handling commonly induces high mechanical load on the lower back of healthcare workers. A long-term research project, the ‘Third Dortmund Lumbar Load Study’ (DOLLY 3), was conducted to investigate the lumbar load of caregivers during handling activities that are considered ‘definitely endangering’ in the context of worker’s compensation procedures. Nine types of handling activities in or at a bed or chair were analysed. Measurement of action forces via specifically developed devices and posture recording by means of optoelectronic marker capturing and video recordings in order to quantify several lumbar-load indicators was previously described in detail. This paper provides the results of laboratory examinations and subsequent biomechanical model calculations focused on lumbar load and the potentials of load reduction by applying biomechanically ‘optimized’ transfer modes instead of a ‘conventional’ technique and, for a subgroup of tasks, the supplementary usage of small aids such as a sliding mat or a glide board. Lumbosacral-disc compressive force may vary considerably with respect to the performed task, the mode of execution, and individual performance. For any activity type, highest values were found for conventional performance, lower ones for the improved transfer mode, and the lowest compressive-force values were gathered when small aids were applied. Statistical significance was verified for 13 of these 17 comparisons. Analysing indicators for asymmetric loading shows that lateral-bending and torsional moments of force at the lumbosacral disc may reach high values, which can be reduced considerably by implementing an improved handling mode. When evaluating biomechanical loads with respect to age- and gender-specific work-design limits, none of the analysed tasks, despite execution mode, resulted in an acceptable load range. Therefore, applying a biomechanically adequate handling mode combined with small aids to lower the friction between patient and surfaces is highly recommended, especially to prevent overload in older caregivers.

Keywords: biomechanical model; healthcare workers; lumbar load; overload prevention; patient handling
INTRODUCTION

Occupational activities involving the handling of heavy objects exert forces that may result in high mechanical load on the lumbar spine (e.g. Bradford and Spurling, 1945; Morris et al., 1961; Chaffin, 1969; McGill and Norman, 1985; van Dieën et al., 2003; Lavender et al., 2009). Besides other influences like psychological and psychosocial, genetic and physiological risk factors (e.g. Frymoyer et al., 1980; White and Panjabi, 1990; Goel et al., 1999; Adams et al., 2002; Schumann et al., 2010), physical work is considered a relevant cause of low-back pain and even severe lumbar diseases when performed long term (e.g. Hoogendoorn et al., 1999; Kuiper et al., 1999; Lötters et al., 2003; Seidler et al., 2009; da Costa and Vieira, 2010).

Low-back disorders are one of the most frequent factors for health-related absenteeism in the workplace (e.g. Liebers and Caiffier, 2009; BKK, 2011; DAK, 2011). This is particularly true for workers in the healthcare professions (e.g. Videman et al., 1984; 2005): Whenever patients, residents, clients or any other individual who receives assistance are moved, high biomechanical low-back load is induced (e.g. Daynard et al., 2001; Skotte and Fallentin, 2008). Therefore, not only nurses but any ‘caregiver’ who is lifting or lowering; pushing or pulling; moving or handling; manoeuvring or holding; carrying or supporting another person on a ward, in an operating theatre, outpatient department, residential or nursing home or similar without adequate guidance or training has a high risk for lumbar complaints or disorders (e.g. Smith, 2005; ISO/TR 12296, 2012).

A couple of national guidelines were published in recent years to assist employers and employees in reducing the risk of overexertion and the incidence or severity of injuries. Although these guidelines for the improvement of healthcare work are available in a few countries like the USA (Feletto and Graze, 2001; Cohen et al., 2010; PSCI, 2005), Canada (WS BC, 2006), UK (Smith, 2005), Australia (WS Vic, 2009) and the Netherlands (Knibbe and Friele, 1999; ISO/TR 12296, 2012), the scientific substantiation of specific advice or regulation is generally not clear. As a consequence, in the past few decades, several research groups have analysed patient-handling activities with regard to the resulting biomechanical load and injury risk in the low-back region. De Looze et al. (1994) investigated the lumbar load during exemplary patient transfer activities, which were considered symmetrically performed, with special emphasis on bed-height adjustments. The investigations of Warming et al. (2009) were based on logbook registrations to record the workload of consecutive days to examine musculoskeletal complaints in diverse body regions.

The Danish research groups led by Skotte and Schibye applied a biomechanical model methodology, partly accompanied by electromyographical measurements and posture recordings to calculate spinal forces and moments of force as typical lumbar-load indicators for patient handlings in or at the bed or chair (Skotte et al., 2002; Schibye et al., 2003; Skotte and Fallentin, 2008). Yassi et al. (2001) explored subjectively recorded results regarding the improvements, which were attributed to the assured availability of mechanical lifters and other assistive devices for reducing physical demands, on work fatigue, back and shoulder pain, and physical discomfort. In a substudy, Daynard et al. (2001) investigated biomechanical outcomes and injury risk; lack of training or non-compliance with applied equipment and handling technique resulted in higher lumbar loading. Nelson et al. (2003) verified improvements by redesigning task performance or using mechanical lifting devices.

In Germany, several patient-handling activities are considered ‘definitely endangering’ with respect to worker’s compensation procedures, in particular when executed in commonly applied ‘traditional’ handling modes. Comparing literature findings and actual demands in German regulation identified deficits regarding the scope of previously analysed tasks or the applied methodology. As a consequence, this study was conceived in order to analyse whether the load on the lumbar spine during specified manual patient-handling activities can be reduced by applying biomechanically improved transfer modes instead of conventional techniques and, exemplarily, by supplementary use of small aids like a sliding mat or glide board.

METHODS

To investigate the biomechanical load on the lumbar spine of healthcare workers during manual patient handling, an intensive research project, the so-called Third Dortmund Lumbar Load Study (DOLLY 3; cf. Jäger et al., 2008) was conducted in cooperation with the German Social Accident Insurance Institution for the Health and Welfare Services (BGW). The examinations were performed in a laboratory setup to enable a comprehensive measurement-based methodology. Two female caregivers with extensive professional experience, as
well as serving as instructors for training supervisors, acted alternately as nurse or patient throughout the study, i.e. no real patients were recruited due to ethical as well as technical reasons. The study was designed to avoid overloading the subjects during patient handling; in particular at high-loading tasks, the patient gave certain support during the handling action. Overall, a patient’s support was qualitatively considered by two grades of mobility as partially or fully cooperative; in other words, the manual handling of a totally non-cooperative person (as when narcotized) or a fully self-sustaining patient remained unexplored.

The methodology to determine a caregiver’s lumbar load quantitatively was described in detail previously (Theilmeier et al., 2010), whereas this paper is focused on the biomechanical low-back load and the potentials for load reduction. Nevertheless, the main issues of both the experimental procedure and the subsequent computations are sketched in the following sub-chapters, supplemented by introducing the scope of analysed tasks.

Procedure overview

Adopted posture and exerted forces are considered the most relevant factors influencing the load on the lumbar spine during manual patient handling; therefore, posture and forces were recorded to ensure a realistic approach. Fig. 1 illustrates the underlying methodology in principle: Data on posture of the nurse (cf. upper-left box) and of the patient (cf. lower left box) as well as data on the forces transmitted from the nurse to the patient in order to initiate, support or perform the intended local shifting of patient’s body or body parts (cf. middle-left box) were continuously obtained during the actions. From this pool, nurse’s data on posture and action forces were separated and subsequently used to estimate relevant indicators of her lumbar load in the temporal course of a transfer action (cf. right side); in particular, the moments and forces with respect to the lowest spinal disc between the 5th lumbar vertebra and the upper part of the sacrum (lumbosacral disc ‘L5-S1’) were quantified considering their vectorial behaviour.

Postural data of the nurse were gathered via common video recording and optoelectronic cameras. The latter, being sensitive in the infrared lightspectrum region, enable detection of light-emitting diodes (LED) attached at relevant anatomical landmarks of the nurse’s body—like the back of the hand, shoulder and ankle joints—as well as at stationary reference points, such as a corner of the bedframe. Three infrared cameras are arranged along a defined distance respectively and in a fixed angle to form a ‘position sensor’. Two of these three-camera sensors were mounted at opposite walls of the laboratory (see Fig. 2). Application of two cooperatively acting position sensors resulted in an accuracy of marker-position determination of about 0.1 mm in a calibrated space of about 5 m length, 2 m width and 2 m height in the centre of the laboratory.

Regarding video recording one camera was installed on the ceiling to preferentially show the nurse’s lateral-bending, turning, and lateral arm positions; one camera was fixed at the laboratory’s sidewall documenting the trunk’s forward inclination, spinal curvature and frontal arm positions; and one camera overviewed the total scene spatially. Video-based data became necessary, in particular, if a marker was temporarily hidden during a handling action so that missing optoelectronic data were supplemented by video-based ones; for example, when the nurse grasps the patient from the underside (cf. Jordan et al., 2011a).

The forces exerted by the nurse were recorded via a specifically established force-sensory bed and chair. A common hospital bed was equipped with a supplementary framework between the bedstead and the bedspring frame, and four tri-axial piezoceramic force sensors inserted at the four corners enabled continuous recording of the reaction forces with regard to magnitude, direction, and distribution in the bed area. As the bedframe forces result from both the caregiver’s action forces and the patient-related forces due to her mass, these influences had to be separated. Therefore, the patient’s posture was recorded via an additional video camera, and subsequent patient-related biomechanical model calculations yielded to corresponding coordinates of the patient’s centre of gravity. As the caregiver applied forces not only to the patient but also to the bed directly, additional force-sensory bars were fixed to the sides of the bed (cf. Fig. 2).

Force splitting was analogously performed when the patient was moved to or picked up from a chair, which was positioned on a common force plate equipped with four 3-axial force sensors at its four corners. The reaction forces at the force plate are a combination of both the caregiver’s exertions and patient-related weight forces superimposed. An additional force plate was used to record the forces under the patient’s feet: when, for example, the reaction forces under the chair disappear due to postural change from a sitting to an upright standing position, the point of application of the weight-related forces change from one to the other force plate. Analogously, force plates in front of the bed were used to control the reaction forces when the
Fig. 1. Schematic representation of the methodology for lumbar-load quantification—Recordings of adopted posture and exerted forces of the nurse serve as input data for biomechanical model calculations to quantify indicators of the nurse’s load on the lumbar spine during manual patient handling; the patient’s posture is recorded to determine body’s position (centre of gravity) related to the respective device involved (bed, chair, floor, force plate).

Fig. 2. Laboratory views—Selected measurement equipment (a) for the recording of the nurse’s postural data via infrared cameras mounted in rigid ‘position sensors’ at opposite walls and (b) for the recording of the nurse’s action forces via a specifically developed force-sensory bed or via ground-reaction force plates under the feet of the nurse (for details, see text).
patient was leaving the bed (cf. Fig. 2; for details, see Theilmeier et al., 2010).

Laboratory views are shown in Fig. 2: the force-sensory bed is located in the centre of the laboratory between the two ‘position sensors’ fixed at opposite walls near the ceiling. Along the bed’s length, stiffening frameworks were mounted to limit bending effects in the bedframe and, thus, to prevent the application of torques on the force transducers at the bedframe corners. Via pedestals of various heights, the vertical position of the caregiver in relation to the bed was adjusted; furthermore, height differences between laboratory floor and force-plate surface were compensated.

**Lumbar-load quantification**

The mechanical load on the lumbar spine was quantified applying a previously developed 3D multi-segmental dynamic simulation tool (‘The Dortmunder’; Jäger et al., 2001a). The underlying biomechanical model is built on inverse dynamics algorithms, i.e. load indicators such as forces and moments of force at the reference point L5-S1 are calculated on the basis of identified posture, movement, and forces acting on the human body. The human skeletal structure is represented by 30 mechanically rigid body segments that are supported in 27 punctiform joints so that common postures and body movements including bending and twisting within the trunk can be replicated realistically.

The human muscular structure is simulated with respect to the region of interest only, here the lower trunk, by a total of 14 muscle cords at the back and abdominal wall, spreading over the lumbar discs and connecting the pelvis and the rib cage biomechanically. The loading moments in the sagittal, coronal, and transversal planes due to posture, motion, and action forces are counteracted by an adequate activation pattern of the nine muscle-equivalent force vectors in the model; as the physiologically based optimisation criterion, the sum of muscle forces is considered minimal so that co-contraction effects are also assumed minimal (cf. Jäger et al., 1991; 2001a).

Here, the lumbar-load estimation for caregivers replicating a manual patient-handling action in the laboratory is computed utilizing the recorded data of the nurse’s motion and action forces of the hands, with consideration to Newton’s mechanics, rules of gravity, inertia and leverage of the body parts. In particular, ‘dynamic’ effects were included so that forces and moments of forces due to translational acceleration and deceleration as well as rotation of body parts were considered in order to quantify lumbar load as realistically as possible.

*Scope of analysed activities*

Activities in or at the bed or chair, analysed in the current study, were chosen due to their classification by the Social Accident Insurance Institution for the Health and Welfare Services (BGW) as ‘definitely endangering’ with respect to the corresponding occupational disease in Germany concerning lumbar degeneration due to manual handling (BMA, 1992). According to a previous examination (cf. Jordan et al., 2011a) these handling activities were selected for which, compared with conventional performance, a biomechanical benefit was assumed by applying an improved technique or, for a subgroup of tasks, by additionally using a commercially available small aid. These activities include the following:

a) Raising a patient from lying to sitting in bed; or vice versa.

b) Elevating a patient from lying to sitting at the bed’s edge; or vice versa.

c) Moving a lying patient towards the head of the bed (nurse at the bed’s long side); [if small aids: anti-slip mat, sliding mat]
d) Moving a lying patient towards the head of the bed (nurse at the head of the bed); [if small aids: anti-slip mat, sliding sheet]
e) Moving a lying patient sidewards in the bed; [if small aids: anti-slip mat, sliding mat]
f) Inclining the head of the bed with a patient lying in it.

g) Positioning or removing a bedpan.

h) Moving a patient seated at the bed’s edge to a chair; or vice versa; [if small aids: glide board, handling belt]
i) Raising a patient from sitting to standing upright; or vice versa.

A typical activity of repositioning the patient in the bed (activity “c” in the above-mentioned list), executed in three modes, is illustrated in Fig. 3 via representative freeze frames of the corresponding video. The initial picture in each row (left pictures) shows the nurse standing at the bed’s long side adopting a ‘reference posture’ (standing upright with elevated forearms) in order to enable infrared-marker recognition and allocation. The patient lies in the bed in a supine posture with the knees pulled up considerably; she is initially positioned near the foot of the bed. At action’s end (right pictures), the patient is repositioned in the bed, which is indicated by a quite smaller flexion angle in the knees. With respect to the median length axis of the bed, she lies about 10 cm closer to the carer in order to avoid an unusual distance, i.e. to compensate for the additional
distance due to the stiffening framework mounted on the bed.

A conventional mode assumed (upper row). The nurse bends her trunk to the side and grasps the patient under her left shoulder region. Here with the right arm from the axillary side and, with her left arm under the patient’s shoulder blade from the acromial side (second picture). In a relatively powerful and jerky movement, the nurse ‘dragged’ the patient towards the head of the bed (3–5 pictures). Thereby, she turns her upper body to her right side, which is accompanied by twisting her trunk. Throughout the action, the patient provided support by partially lifting her body, especially her pelvis, via pulling up at a monkey pole with her arms and by pushing her body by repelling with her feet.

The second sequence of pictures (middle row) corresponds to a biomechanically ‘optimized’ mode of exertion, according to the ‘BGW concept’ (Baum et al., 2012; Fischer, 2010). Prior to the main action, the nurse performs some pre-positionings (not visualized here): turning the patient a little away enables the nurse to put her right forearm under the patient’s left part of the trunk so that the trunk will glide like on a skid. In addition, the nurse positions her left forearm at the patient’s crossed arms (over the ribcage) (second picture) so that, subsequently, the force may be effectively transmitted (3–5 pictures). Throughout the action, the patient provided support by repelling with her feet at the heels. Thereby, the upper bodies of nurse and patient build ‘a compact unit’ where relative motion among one another is very limited: the caregiver’s posture is characterized by stiffening her upper body from the hips to the arms and laterally moving her pelvis by shifting her weight from the left to the right foot.

The lower row of pictures illustrates the third execution mode when small aids are applied. Equipment positioning was performed prior to the situations demonstrated in the figure; corresponding low-back load is provided in a previous paper (Jäger et al., 2008). In this case, a sliding mat was positioned under the patient’s trunk in order to lower the friction...
between the body and the sheet on the mattress. Furthermore, an anti-slip mat was put under the patient’s feet to avoid slippage, i.e. to increase friction between feet and mattress and, hence, to enable the patient to support by repelling with her feet. The second picture corresponds to that point in time when the nurse had completed positioning her hands and arms: she laid her right forearm under the patient’s shoulder and neck in order to support the patient’s head, whereas her left forearm was positioned at the patient’s thighs near the buttocks to build a proper bearing. Afterwards (3–5 pictures), the repositioning task is performed similar to the execution explained in the previous paragraph via a ‘smooth’ movement of the compacted upper bodies of nurse and patient.

RESULTS

Exemplary time courses

Fig. 4 shows exemplary time courses of about 10-s length resulting from the laboratory examinations regarding the activity explained in Fig. 3: repositioning a patient in bed. For clarity reasons, the time courses were divided into six phases: standing upright in the initial reference posture; leaning forward and/or sideward towards the patient; performing some pre-positionings; the main task of ‘moving the patient’; re-erecting the upper body; changing posture into a final upright standing position. The initial and final upright-standing phases indicate reference values quantified via biomechanical model calculations prior to the laboratory measurements. No data were obtained in the re-erecting phases for optimized task execution, which led to interruptions in the curves due to time restrictions during data capturing.

The upper row of time courses corresponds to the sagittal, lateral, and vertical components of the action forces, each representing the resultant for both hands. The traces in the middle row show the components of the moments with respect to the nurse’s lumbosacral disc. The bottom traces demonstrate the components of the reaction forces at the lumbosacral
joint. Besides the compression component, the shear forces in the anterior-posterior direction (‘sagittal’) and those pointing to the side (‘lateral’) are drawn. Lateral shear at L5-S1 amounts to zero in the initial and final phases of the task, when the nurse adopts a symmetrical upright position without any action-force exertion. In contrast, both the compressive and the sagittal shear forces at the lumbosacral disc are unequal to zero in the upright posture due to weight-induced forces of the body parts superior to the reference disc L5-S1 and its ventral inclination by ~30° through normal spinal curvature. According to coordinate convention, the reaction force component ‘sagittal shear’ adopts negative values here.

The highest action forces at the hands, loading moments, and reaction forces at the spine are generated in the ‘moving’ phase, whereas, albeit showing lower values, considerable forces and moments are also found in the time sections of pre-positioning and grasping the patient. Moving the patient in a conventional way by grasping under the shoulder and ‘dragging’ her to the right side in a relatively jerky movement yielded an upward directed action-force peak of about 300 N vertically, followed by a large horizontal action force of about 350 N, with a negative value due to coordinate convention (cf. moving section in upper-left diagram). These action-force peaks correspond to distinct extremes in the moment-component courses (middle-left diagram). The sagittal moment reaches more than 300 Nm according to the lifting phase, whereas the torsional peak amounts to nearly 200 Nm and the lateral-bending moment to more than 100 Nm while shifting the patient horizontally to the right side. The corresponding lumbosacral reaction forces also show distinct extreme values: the compressive component amounts to about 6.5 kN, whereas shear reaches nearly 1 kN sagittally and 0.6 kN laterally.

Comparison of corresponding values for the optimized mode, without and with application of small aids, shows smaller values for most indicators. For example, the vertical action-force component is diminished from 300 N to less than 50 N (optimized) or to negligible amount (optimized + small aids); analogously, the horizontal component to the right side is decreased from about 350 over 300 to 100 N. The moment components are characterized by a similar, but not identical behaviour: The sagittal bending moment is up to 300 Nm when the task is conventionally performed, about 150 Nm for an optimized execution, and less than 100 Nm when applying small aids; in contrast, the peaks of the lateral bending and torsional moment components are in the same order of magnitude for the conventional and the optimized mode (around 150 Nm), whereas they are substantially decreased through the application of small aids (about 50 Nm). In addition, the calculated disc forces are also diminished for an optimized handling technique: the peak is lowered from 6.5 over 5.5 to 2.5 kN, whereas sagittal and lateral shear peaks are almost the same for conventional and optimized performance, but more than halved if small aids are applied.

**Lumbosacral compressive forces**

Fig. 5 demonstrates the aggregation of findings for nearly 140 actions, which are categorized into nine handling activities, each of which was performed in two or three modes so that results for 22 groups of actions are provided in all. The diagram shows that the disc-compressive force may differ considerably with respect to the achieved task, the mode of execution, and in particular, individual performance. A conventional mode assumed, the averaged peaks range between about 3.5 kN for ‘raising from lying to sitting’ in bed (cf. column most left, activity ‘a’) and 6.5 kN for ‘moving the patient to the head of the bed with the nurse acting from the bed’s long side’ (cf. left column for activity ‘c’). With an optimized handling mode, but without the use of small aids, the corresponding range is approximately 2.5–5.5 kN, whereas with the use of small aids, the averaged peaks amount to 2–3 kN. Furthermore, the column pattern shows the highest means for conventional task execution for all nine activities under study, lower ones for an optimized handling mode, and the lowest mean values when small aids are applied. Large differences of about 2 kN or more were obtained for four activities (‘b’, ‘d’, ‘e’, ‘i’) using an optimized rather than a conventional execution mode, whereas force differences of about 1 kN were found for four other activities (‘a’, ‘c’, ‘g’, ‘h’), and a smaller improvement was achieved for one activity only (‘f’: inclining the head of the bed). In this case, the limited load reduction can be attributed to a speciality of task execution, namely pulling a lever positioned under the bedframe enabling a hydraulic support to raise the head of the bed, which entails similar postures for both execution modes.

Statistical significance of different compressive forces for varied execution mode was found for 13 of a total of 17 examinations based on the Student’s t-test. Compared with conventional performance, significantly lower forces were substantiated in eight of nine tested handling activities utilizing an optimized execution mode and in all of the four activities involving the additional use of small aids. Comparing the two optimized techniques, the lumbar-load
reductions differ considerably: For repositioning the patient in bed from the bed’s long side (activity ‘c’), statistical significance was verified, whereas for moving the patient in the bed sideways (‘e’) and moving her from bed’s edge to a chair (‘h’) significance was not reached. Negligible load reduction was found, however, for the patient’s repositioning when acting from the head of the bed (‘d’): this can be traced back to the fact that a large reduction was still achieved by applying an optimized technique, instead of the conventional mode, which was accompanied by dismantling the headboard of the bed and, thus, enabling a much better force exertion: horizontal pulling instead of pulling superimposed with a large lifting component. Furthermore, as specifically related analyses had shown, the patient gave more support when no sliding sheet was used anticipating a high-loading task, whereas in the case of applying the friction-reducing device the patient had a less active role.

Side-bending and torsion

The load on the musculoskeletal system is commonly not symmetrical with respect to the median sagittal plane due to twisting or bending the trunk sideways, lateral arm positions, unequal force exertions of both arms or action-force components pointing laterally. Therefore, further indicators besides disc-compressive force may illustrate the level of asymmetry in such tasks and the complexity of biomechanical spinal loading during manual patient handling. In this context, Fig. 6 shows results for the nine analysed types of activity (‘a’–‘i’) by means of three indicators, i.e. the moment components of lateral-bending (abscissae to the left) and torsion (abscissae to the right) related to the compressive force drawn at the ordinates. The resulting columns for lateral and torsional moments were set at those points at the ordinate which match with the corresponding mean for lumbosacral compression (cf. Fig. 5). In order to maintain clarity, the ranges for the diverse activities and modes are listed in Table 1.

At first sight, the diagrams of Fig. 6 show again that the compressive forces are higher for the conventional mode (~3.5–6.5 kN) than for the optimized mode (~2.5–5.5 kN) and the supplementary use of small aids (~2–3 kN). The corresponding range pattern of the lateral-bending moment (columns to the left) is similar, in so far as lower values are achieved for small-aid application (~15–50 Nm) than for the optimized mode (~20–120 Nm), which in turn has smaller values than the conventional handling mode (~40–110 Nm). Looking to the torsional moments (columns to the right) reveals the highest means for conventionally performed actions (~30–160 Nm), slightly lower ones for optimized task performance (~10–150 Nm) and generally, the lowest values for additional use of small aids.
Upon further evaluation the diagrams show that the moment values increase with increasing compressive force; the highest lateral and torsional moments are combined with the highest compressive forces (cf. left and middle diagrams), and the lowest moments can be found in the lower region of compressive-force values (cf. all diagrams). However, tasks with almost identical compressive-force values may also be combined with substantially different asymmetry-indicating moments. For example, comparing ‘elevating a patient from lying to sitting at the edge of the bed’ (‘e’), and ‘raising a patient from sitting to standing upright’ (‘i’) show similar compressive forces of about 5 kN, if a conventional execution is assumed, but the torsional moment may differ by a factor of three: 102 versus 36 Nm. A similar relation can be found for an optimized performance when examining the results for lateral-bending moments: raising from lying to sitting (‘a’) yielded a mean of 47 Nm, whereas moving a lying patient sidewards in the bed (‘e’) resulted in 17 Nm; the compressive-force average amounts to ~2.5 kN for both. A final example reveals a factor of more than four: This was achieved for repositioning the patient with the nurse acting from the head of the bed (‘d’) in relation to positioning a bedpan (‘g’), if an optimized mode without aids application is performed and torsional moment-means are compared (50 versus 11 Nm); the respective compressive-force means amount to about 2.5 kN for both activities. This illustrates that describing lumbar load by only one indicator may be limited, in particular, if the tasks manifest a high degree of asymmetry in posture, movement or force exertion.

**Discussion**

With the combined approach of motion capturing, action-force measurement, and biomechanical model calculations, estimation of several lumbar-load
Table 1. Overview of lumbar-load values for manual patient handling—means and ranges of three indicators for nine activity types and three execution modes (number of executions per table cell: see Fig. 5).

<table>
<thead>
<tr>
<th>Activity</th>
<th>Lumbar load Mean (range) related to L5–S1</th>
<th>Lateral moment in Nm</th>
<th>Torsional moment in Nm</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Compressive force in kN</td>
<td>Conventional Optimized Optimized + small aids</td>
<td>Conventional Optimized Optimized + small aids</td>
</tr>
<tr>
<td>a. Raising from lying to sitting</td>
<td>3.4 (1.8–5.4) 2.3 (1.9–2.9) n.a.</td>
<td>63 (21–99) 47 (12–82) n.a.</td>
<td>51 (30–90) 42 (23–62) n.a.</td>
</tr>
<tr>
<td>b. Elevating to sitting at bed’s edge</td>
<td>5.0 (3.3–6.2) 2.7 (2.0–3.6) n.a.</td>
<td>76 (40–105) 34 (23–50) n.a.</td>
<td>102 (73–160) 48 (40–65) n.a.</td>
</tr>
<tr>
<td>c. Moving to bed’s head (nurse at bed’s long side)</td>
<td>6.7 (5.6–8.0) 5.4 (3.7–6.5) 2.8 (2.3–3.2)</td>
<td>111 (31–268) 124 (41–192) 37 (15–54)</td>
<td>156 (117–253) 154 (106–223) 55 (42–62)</td>
</tr>
<tr>
<td>d. Moving to bed’s head (nurse at bed’s head)</td>
<td>5.7 (2.8–8.9) 2.5 (2.0–3.0) 2.4 (2.2–2.8)</td>
<td>50 (19–134) 43 (33–55) 18 (8–35)</td>
<td>77 (20–184) 50 (12–65) 34 (12–61)</td>
</tr>
<tr>
<td>e. Moving sideways</td>
<td>4.9 (3.3–5.8) 2.6 (2.0–3.4) 1.9 (1.6–2.2)</td>
<td>49 (33–78) 17 (10–23) 15 (7–27)</td>
<td>55 (40–67) 36 (28–48) 25 (16–39)</td>
</tr>
<tr>
<td>f. Inclining the bed’s head</td>
<td>4.3 (3.8–5.4) 4.1 (3.5–5.2) n.a.</td>
<td>56 (25–78) 36 (25–46) n.a.</td>
<td>58 (27–73) 50 (33–70) n.a.</td>
</tr>
<tr>
<td>g. Shoving the bed-pan</td>
<td>4.2 (2.6–6.5) 2.6 (1.6–3.3) n.a.</td>
<td>36 (21–47) 18 (12–23) n.a.</td>
<td>32 (15–46) 11 (9–14) n.a.</td>
</tr>
<tr>
<td>h. Placing from bed in chair</td>
<td>5.1 (3.8–6.5) 3.7 (2.3–4.4) 3.1 (1.6–5.3)</td>
<td>59 (31–105) 37 (25–48) 24 (8–36)</td>
<td>51 (22–84) 40 (15–56) 38 (19–56)</td>
</tr>
<tr>
<td>i. Raising from sitting to upright</td>
<td>4.9 (3.8–6.4) 2.5 (1.9–3.1) n.a.</td>
<td>40 (30–56) 20 (15–24) n.a.</td>
<td>36 (31–38) 21 (18–25) n.a.</td>
</tr>
</tbody>
</table>

n.a. = not analysed or not applicable.
indicators including compressive forces at the lumbosacral disc of caregivers during manual patient handling was possible. For all nine transfer types analysed, lumbar load was high, which was considerably reduced for eight activities, when the activity was carried out in an optimized mode or with the use of small aids.

**Criticism of the methods**

Posture recording was based on optoelectronic marker capturing supplemented by common video documentation, in particular, to enable data restoration for a temporarily hidden marker. This aspect was present for the initial phase during 5 of the 22 action groups analysed but was limited to one marker at one hand only when the caregiver held her hand underneath a body part of the patient; for example, such grasping of the patient from the underside occurred when raising the patient from lying to sitting in bed. Commonly, these phases of a hidden marker lasted less than about a quarter of a second; maximally, it took up to approximately a total of 1 s. To overcome this problem, using a split-screen technique with simultaneous views from above, laterally and from the opposite side combined with a spatial view, the caregiver’s posture and movement could be replicated in all cases via an iterative procedure. Posture was initially described roughly by a stick figure serving as input for a first digitized representation. The coordinates of the respective ‘hidden’ body segment were then set to coincide with the position of the corresponding marker, and repeating this process, the deviations were minimized (Jordan et al., 2011a).

The caregiver’s action forces were derived from the reaction forces, and their spatial distribution was measured with specific measuring systems. A couple of testing and calibration examinations were performed with respect to linearity, repetition accuracy, intra- and inter-sensor crosstalk, time-related drift of force sensors, and the entire measuring system. Consecutive examinations revealed a high reliability of the system (deviation <5%). The local deviation of a computed point of force application at the ‘measuring bed’ compared with the real position yielded a difference of <2% on average (maximum: <5%; cf. Theilmeier et al., 2010).

Mechanical lumbar-load indicators were quantified from biomechanical model calculations (‘The Dortmunder’; cf. Jäger et al., 2001a). This simulation tool utilizes a relatively comprehensive replication of the relevant skeletal and muscular structures so that several approaches were applied for validation. For example, electromyographical measurements were used to identify the best-fitting positions of the back muscles in relation to the spine so that muscle activity can sufficiently be replicated by muscle-equivalent force vectors. Furthermore, model calculations were compared with intradiscal-pressure measurements provided by Nachemson (1966), Andersson et al. (1977), and Wilke et al. (1999) for several postures and object handlings. Overall, calculated and measured pressure values have shown a good correspondence and in particular, analogous behaviour with respect to posture and handled objects (cf. Jäger, 2001).

Sample size of the provided laboratory measurements may be considered a critical issue. Only two persons served alternately as caregiver and patient, and the patient’s support level varied between partially and fully cooperating. This implies, for example, that the variety of patient behaviour, including non-cooperating due to narcosis or unwillingness, is not completely reflected in this study. The unwillingness or inability to cooperate was excluded as two instructors with extensive experience in training supervisors acted as nurse and patient, and they operated in well-timed coordination according to their experience in team work. Even unintentional, subliminal cooperation, and a supportive behaviour may have been present during the measurements, in particular, when the patient-imitating subject anticipated a transfer inducing high load for the task-performing colleague. This effect was clearly identified, for example, for repositioning the patient when acting at the head of the bed: in order to avoid biomechanical overload for the caregiver, the ‘patient’ was rather more helpful if no friction-reducing sliding sheet was applied in comparison to trials with its usage—although the grade of mobility should be the same (‘partially cooperative’). Furthermore, subject weight was within the usual range (~60 and 80 kg) so that bariatric persons and their load on the spine of the carer was not considered; this topic was studied separately (Jordan et al., 2011b). Consequently, the lumbar load under real circumstances may be higher than provided here.

Small sample size may also be related to the limited number of repeated task executions with more or less identical conditions. Restricting the analysis of ‘typical’ task executions only with respect to the resulting biomechanical load on the lumbar spine seemed reasonable due to the huge amount of time necessary for data evaluation. This can be attributed to the high demands of a spatial recording of posture and movement of the caregiver and patient, combined with a 3D action-force recording and subsequent 3D dynamic lumbar-load description via several indicators. Selection of typical trials of task execution was
possible by means of a five-times larger pool of collected data on postural and action-force related variables. As mentioned above, limited coordination of the two people involved during the handling action, which may lead to competing force exertion and, thus, to higher lumbar load than provided here, was not considered in this investigation.

The number of actions varies and ranges from four executions at the very minimum, up to 16, with a mean of 7.2. The reason for so few repetitions is due to the time-consuming data analysis, as mentioned above, whereas the higher numbers are attributed to several reasons: Diverse alternatives were tested so that, for example, 16 ‘repetitions’ were analysed using two types of aids when placing a patient from the bed’s edge to a chair, i.e. a glide board and a handling belt. Or when repositioning the patient with the carer acting at the head of the bed, one version was performed with a dismantled headboard, the other one with the headboard in place. Testing of two variants was also true for positioning the bedpan. For other activities, however, higher numbers of execution were substantiated by the results: Primary analyses revealed high load values (e.g. moving the patient to the bed’s head with the caregiver acting at bed’s long side) or large ranges (e.g. raising a patient from lying to sitting in bed) so that justification by further analyses became reasonable.

**Comparison with previous studies**

Nelson *et al.* (2003) report on a very comprehensive laboratory examination which included electromyographical recordings, measurements of the ‘external’ forces and subsequent biomechanical model calculations for predicting the joint moments and forces acting at the lumbar spine. The authors were able to identify high risks for several handling tasks and recommended redesign. Unfortunately, the argument for identifying a biomechanical benefit is not supported by load data with regard to the spine, instead, it is limited to exemplary values of relative reduction by adopting improved postures or through the use of lifting devices. Furthermore, the undoubtedly promising study was performed by means of manually handling a manikin instead of a human subject. Due to ethical and safety reasons, McGill and Kavcic (2005) analysed a manual ‘patient’ bed-to-bed handling activity with respect to friction reducing assistive devices and the resultant low-back load by means of a patient- replicating manikin too. Substantiating the supposed load reduction in principle, the large influence of handling technique and differences in the properties of devices were pointed out. Regarding the latter aspect, Connolly *et al.* (2001) have found not only small differences in the mechanical effect of diverse slides but also differences in monetary costs.

Comparable limitations were found for studies where patient properties were considered in a rough manner only. Garg *et al.* (1991a,b) applied a biomechanical model for spinal-force prediction which neglected inertial effects, whereas Owen *et al.* (1992) estimated the action forces exerted during a transfer of half the patient’s weight without considering horizontal force components, whereas in the investigation of Marras *et al.* (1999) a person with a relatively low body mass of 50 kg served as a ‘standard patient’. In more recent investigations, Garg (2006) focused on the prevention of injuries in nursing homes and hospitals by implementing a ‘zero-lift program’ in the form of a participatory team approach, without evaluating the resultant biomechanical load. Marras *et al.* (2009) investigated two types of transfer devices in a laboratory, a ceiling-based and a floor-based patient lifter, with respect to lumbar-load indicators. In this context, Chhokar *et al.* (2005) identified economic benefits of applying a ceiling lifter when patients are to be lifted, transferred or repositioned. Comparison shows that the topics of those examinations deviate considerably from the current study.

A more similar approach of combined motion capturing, action-force measurement and biomechanical model calculations was applied by the Danish research group of Skotte, Schibye and coworkers. Skotte *et al.* (2002) provided investigations into low-back load for six types of activity, performed as during normal work, that is to say, without any actual instructions or restrictions during the investigation. The highest compressive-force values were found for tasks involving lifting heavy body parts of the patient (~4–4.5 kN), intermediate values for repositioning tasks in the bed (~3 kN) and the lowest for turning a lying person in the bed (~1.5–2 kN). This phenomenon was also established in the present study: tasks combined with lifting elements lead to the highest compressive forces (~3.5–6.5 kN), if a ‘conventional’ handling mode is assumed. Lower values were adopted (~2.5–3 kN) when lifting components were minimized in an ‘optimized’ execution; turning a lying patient in the bed was not analysed here.

The considerably higher values for the activities implementing lifting in the current study can be attributed to the fact that here only those actions that were considered as putatively linked with high loading and overload risk for the lumbar spine with respect to the corresponding German occupational disease were analysed. Consequentially, the scope of analysed tasks differed essentially between the two
A distinct reduction of ~2 to 3 kN of disc-compressive forces was observed due to mode variation in both the Schibye examination and the present study. Another possible explanation may be differences in execution technique or conditions of the task performed, as Skotte et al. classified this task as ‘non-lifting’ whereas under our examination lifting components in the action force adopted generally high peaks ranging between 230 and 400 N; this fundamental difference suggests that the activity was quite different in the two studies despite a similar description. Based on the results of Skotte et al., Schibye et al. (2003) reported on the effect of changing the patient-handling technique with respect to lumbar load, in part, by applying friction-reducing small aids under the pelvis as well as friction-increasing auxiliaries under the feet of the patient. A distinct reduction of ~2 to 3 kN of disc-compressive forces was observed due to mode variation in the head of the bed with the nurse acting at the bed’s long side: Skotte et al. provide a mean compression of 3.1 kN whereas in the present study a value of 6.7 kN was calculated. Another possible explanation may be differences in execution technique or conditions of the task performed, as Skotte et al. classified this task as ‘non-lifting’ whereas under our examination lifting components in the action force adopted generally high peaks ranging between 230 and 400 N; this fundamental difference suggests that the activity was quite different in the two studies despite a similar description. Based on the results of Skotte et al., Schibye et al. (2003) reported on the effect of changing the patient-handling technique with respect to lumbar load, in part, by applying friction-reducing small aids under the pelvis as well as friction-increasing auxiliaries under the feet of the patient. A distinct reduction of ~2 to 3 kN of disc-compressive forces was observed due to mode variation in both the Schibye examination and the present study.

**Overload risk**

To emphasize the importance of low-back load and overload for nursing personnel, injury rates and absenteeism data are often cited in the literature (e.g. Menzel, 2004; Videman et al., 2005; Nelson et al. 2008). Besides the load induced by the large number and long duration of trunk-flexed postures adopted during the normal working day (Freitag et al., 2007a; 2012), manual patient handling is frequently considered a high-risk task (e.g. Garg et al., 1991a; Schibye et al., 2003; Waters, 2007). To find a more distinct relation between work and the justification of overload, calculated disc-compressive forces were compared with spinal load-bearing capacity in order to form a biomechanically oriented assessment of lumbar load.

The well-known NIOSH method (NIOSH, 1981; Waters et al., 1993) is frequently used for the evaluation of manual lifting activities in ergonomics and occupational health. This approach established the biomechanical criterion of 3.4 kN as the upper limit for lumbosacral-disc compressive forces based on experiments for determining the ultimate compressive strength of cadaveric lumbar-spine sections. As ‘in terms of a specification for design a much lower level should be viewed as an upper limit’ and as ‘this will not necessarily be protective for most individuals over 50 years of age or other susceptible populations’ (NIOSH, 1981), this research of biomechanical load-bearing capacity of the lumbar spine was continued. In this context, the so-called Dortmund Recommendations (Jäger et al., 2001b; ISO 11228-2, 2007) were derived from a relatively large database, considering age and gender as the most influential factors on failure load. A higher biomechanical load-bearing capacity among males than females and a steeper decrease over age for males than for females were found. According to the high proportion of women in the healthcare professions, the recommended range of 4.4 kN for young women (20 years) and 1.8 kN for older caregivers (≥60 years) is appropriate while handling patients manually.

Considering these criteria, none of the analysed tasks, despite the mode of execution, resulted in load values lower than the lowest limit recommended for older females. Therefore, manual patient handling under the given conditions might be interpreted as a certain biomechanical overload risk for the lumbar spine of the performing caregiver. Furthermore, several activities resulted in disc-compressive forces exceeding the highest recommended limit for women. Hence, such handlings must be regarded as unacceptable for all women. Even young female adults should not work under the respective conditions, and an optimized handling mode should be established and put into practice.

Essential load reduction is achieved when additional small aids are used. Even the task associated with the highest disc-compressive forces in the study, i.e. moving the patient towards the head of the bed, can be performed with a significantly reduced overload risk for caregivers aged 40 years or older: either the healthcare worker acts at the head of the bed to enable a more symmetric posture, in which case the carer should dismantle the headboard to render confining the vertical force of exertion for ‘lifting’ components possible, and so she can pull the patient towards her own body while keeping lateral and vertical forces of exertion small. Or, alternatively, the healthcare worker acts at the long side of the bed to perform a lateral transfer. Here friction affecting devices are necessarily to limit the lateral action forces; to diminish asymmetry in posture, the carer should shift her weight sideways from one foot to the other while stiffening her upper body and establishing a compact unit with the patient’s body during the movement. Then, disc-compressive forces are kept within a reasonable range, and the asymmetry-indicating load parameters of lateral and torsional moments are significantly decreased as well.
Finally, it should not be ignored that task execution is prolonged when applying a biomechanically optimized handling mode or small aids. However, the frequency of manual patient-handling actions per shift is rather small so that the accumulated additional time is relatively short. As previous examinations of the German Social Accident Insurance Institution for the Health and Welfare Services have shown, ~55–70 handling actions are performed during a shift, less often on hospital wards and more often in home care and nursing homes (BGW, 1995). More recent shift data (Freitag et al., 2007b) revealed lower frequencies of handling of 16–20 per person per shift lasting an accumulated total of about 2 min. Assuming roughly 100% additional time for applying an improved handling mode demonstrates the negligible amount of time in comparison to the high biomechanical benefit due to the reduction of lumbar-overload risk.

CONCLUSIONS

The results of the ‘Third Dortmund Lumbar Load Study’ (DOLLY 3) confirm previous literature findings that the biomechanical load on the lumbar spine through manual patient handling should not be characterized as non-hazardous, in contrast, lumbar load may be high or even too high for caregiving personnel. This is induced, in part, by high action forces exerted to position or reposition the patient in the bed or on a chair, or by force exertion in unfavourable postures. High lumbar load causes large disc-compressive forces as well as in intensive asymmetric loading like bending and torsional moments of force. Those torque components—which should be minimized when possible—result from lateral force exertion for sideward moving or from bilaterally different hand-forces exerted during turning a patient, for example.

As a considerable overload risk was assumed for a couple of tasks, which was categorized, prior to the present study, as ‘definitely endangering’ in German worker’s compensation procedures, solutions for eliminating work-design deficits were intensively sought. DOLLY 3 substantiated the hypothesized biomechanical benefits of ‘improved’ handling modes over ‘conventional’ executions for several handling actions in or at the bed or chair. In addition, the application of small aids like a sliding sheet or glide board leads to further lumbar-load reduction during the performance of certain examined activities associated with the highest lumbar load, i.e. the hypothesis of a positive effect of such devices was confirmed by specific experimental findings. In conclusion, recommendations for applying biomechanically optimized handling modes supplemented by small auxiliaries are substantiated and make these preventive measures reasonable to reduce the biomechanical load on the caregiver’s spine and, hence, to achieve a necessary reduction in lumbar-overload risk for the working staff in the healthcare professions.

FUNDING

German Social Accident Insurance Institution for the Health and Welfare Services (Berufsgenossenschaft für Gesundheitsdienst und Wohlfahrtspflege, Hamburg, Germany)

Acknowledgements—Special thanks for the very helpful support and advice of Mrs B.B. Beck and Mrs B. Wiedmann (BB, Hamburg, Germany) who, due to their extensive experience in the field of nursing, were superlatively capable of performing the patient-handling tasks fundamental to this study. Furthermore, the authors would like to express heartfelt thanks to Mrs M. Sandusky (IfADo, Dortmund, Germany) for very carefully editing the English of the paper on hand. Finally, the authors want to thank the members of the ‘Dortmund Lumbar Load Study Group’ for the very fruitful and intensive collaboration; members are R. Göllner, Y. Güler-Öztürk, M. Konhoff, K. Lukaszewski, J. Mattern, J. Metzler, M. Mujih, U. Öztürk, M. Schmitz, T. Scholz, R. Smolka, S. van der Wel, J. Voß, F. Ziolkewicz and the IfADo authors.

REFERENCES


Lumbar-load analysis of manual patient-handling activities


