Measurement of Action Forces and Posture to Determine the Lumbar Load of Healthcare Workers During Care Activities with Patient Transfers

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Moving patients or other care activities with manual patient handling is characterized by high mechanical load on the lumbar spine of healthcare workers (HCWs). During the patient transfer activity, the caregivers exert lifting, pulling, and pushing forces varying over time with respect to amplitude and direction. Furthermore, the caregivers distinctly change their posture and frequently obtain postures asymmetrical to the median sagittal plane, including lateral bending and turning the trunk. This paper describes a procedure to determine lumbar load during patient transfer supported by measurement techniques and an exemplary application; this methodology represents the basis of a complex research project, the third ‘Dortmund Lumbar Load Study (DOLLY 3)’. Lumbar load was determined by simulation calculations using a comprehensive biomechanical model (‘The Dortmunder’). As the main influencing factors, the hand forces of the caregiver exerted during typical patient transfers and the posture and movements of the HCW were recorded in laboratory studies. The action forces were determined three-dimensionally with the help of a newly developed ‘measuring bed’, two different ‘measuring chairs’, a ‘measuring bathtub’, and a ‘measuring floor’. To capture the forces during transfers in or at the bed, a common hospital bed was equipped with an additional framework, which is attached to the bedstead and connected to the bedspring frame via three-axial force sensors at the four corners. The other measuring systems were constructed similarly. Body movements were recorded using three-dimensional optoelectronic recording tools and video recordings. The posture and force data served as input data for the quantification of various lumbar-load indicators.

Keywords: action forces; back pain; biomechanical model; health-care workers; lumbar load; manual handling; nurses; posture

INTRODUCTION

Manual materials handling activities are connected with a high risk for the development of diseases related to the intervertebral discs (Videman et al., 1984; Luttmann et al., 1988; Riihimäki et al., 1989; Hofmann et al., 1995; Hofmann and Korn, 2001; Seidler et al., 2009, 2003). Furthermore, diseases of the muscle and skeleton systems are one of the most frequent causes for health-related absenteeism in the workplace (European Communities, 2002; BKK-Bundesverband der Betriebskrankenkassen, 2008). Similarly, occupational low back pain is a significant problem among nurses (Mitchell et al., 2009) and care activities with patient transfer may lead to high load on the spine and may accelerate the development of degenerative disc-related diseases in the long run of occupational life (Videman et al., 2005).

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of Technology (IfADo), in cooperation with the Statutory Accident and Health Insurance Institution for Health Services and Welfare Care (BGW), the ‘Dortmund Lumbar Load Study 3 (DOLLY 3)’ was conducted (Jäger et al., 2007; Theilmeier et al., 2010). The study represents a long-time research project on the determination of lumbar load in selected care activities with patient transfers. It was accomplished to assess lumbar diseases with respect to mechanical loads in occupational fields and for the development of preventive procedures for the avoidance of lumbar spine’s disease. This paper describes the first part of the study and is primarily concerned with the methodology to determine the lumbar load of healthcare workers (HCWs) performing patient transfers supported by measurement techniques and an exemplary application.

When moving patients during care activities, HCWs frequently adopt postures asymmetrical to the median sagittal plane, including lateral bending and turning the trunk, and laterally positioned arms including sideward force exertion. Furthermore, they exert lifting, pulling, and pushing forces varying over time with respect to amplitude and direction. Therefore, a three-dimensional (3D) determination and replication of both, the posture and the action forces in high temporal resolution, are needed for an adequate approach for quantifying biomechanical indicators of load on the lumbar spine (Jordan et al., 2006; Theilmeier, 2006).

In the research project, the quantification of the lumbar load was performed by model calculations using a previously developed simulation tool (‘The Dortmunder’; Jäger et al., 2001a). The calculation of mechanical lumbar-load indicators with this 3D multisegmental dynamic biomechanical model presupposes knowledge; firstly, on forces acting on the nursing person’s body, commonly applied at the hands (so-called action forces) and, secondly, of caregiver’s posture.

With a similar approach in former studies, Garg et al. (1991) as well as Owen et al. (1992) examined the biomechanical load on the lumbar spine with a static biomechanical model, whereby they assumed that the forces at the hands of the HCW correspond to the half patient weight. Furthermore, in their model, lumbar-load computations were only possible for vertical action forces, i.e. forces acting spatially could not be considered. In a similar way, Zweiling (1996) accomplished the biomechanical computations with a comparatively simply structured ‘three-angle model’. This permits analyses of symmetrical slow lifting operations in the sagittal plane only. By recording vertical forces with measuring soles under the feet of the HCW, Morlock et al. examined care activities; in consequence of this procedure, horizontal forces on the worker could not be considered (Morlock et al., 1997).

Deuretzbacher and Rehder (1997) used a biomechanical model, which represents the musculature in the lower back by two force resultants acting in parallel to the spine. This very limited muscular structure modelling permits adequate computations for activities without lateral force components only. Because of the limitations of the described methods, the new technology—described in this paper—and was developed aiming for a more detailed and close-to-reality description of lumbar load during patient transfer activities.

**METHODS**

The posture and the forces exerted by the HCW are the most relevant factors concerning the lumbar load of a person. The data are needed as input to calculate the lumbar load with the biomechanical model. Following the aim of a detailed determination of the lumbar load presupposes knowledge of the time courses of the action forces and the posture 3D and in high time resolution. To this end, posture and force data were recorded with a sample rate of 100 Hz. The methodology used for the studies requires high personnel effort for gathering and evaluation of the data, and therefore, it is less suited for larger groups of subjects. The disadvantages of evaluating a relatively low number of task executions could be reduced by a careful selection of typical, i.e. ‘average’ tasks from the pool of the totally collected data and by utilizing the help of two highly experienced caregivers acting alternately as a ‘typical’ HCW or a typical patient.

**Posture recording**

The posture and the movements of the nursing personnel were gathered with the help of a combination of two measuring systems: video analysis and optoelectronic measurements. Figure 1 gives an overview of the laboratory and the equipment.

The posture was captured via video documentation from different lines of vision with four video cameras (Video 1 up to Video 4). Additionally, a 3D optoelectronic motion capturing system was used to track continuously the coordinates of small infrared markers (diameter 10 mm, active area <1 mm) attached to relevant body parts of the subject (shoulder, hand, hip, and heel, each at the right- and left-hand side) and, for reference, at the bed frame. The markers’ spatial localizations were recorded
via two ‘position sensors’ consisting of three infrared cameras each. The position sensors were mounted at opposite walls of the laboratory. The measuring space of the position sensors overlaps and so a common measuring area is formed enabling the tracking of the markers at both sides of the body of the subject. The aperture angle of the optical systems leads to a hexagonal measuring area in the top view (coloured grey in Fig. 1; length \( \sim 5.0 \) m, width between 1.2 and 1.6 m, and height 2.0 m). The position sensors act cooperatively and, therefore, a changeover of a marker from the measuring area of the first sensor to the measuring area of the second sensor was possible. So failures in the detection of the markers (e.g. due to covering of the markers by the patient) could be significantly reduced.

Applying a graphical animation system, the real posture of the nurse was reproduced digitally in an iterative procedure by combining the video and the optoelectronic data. Posture data were transformed into a digital format suitable for the subsequent lumbar-load quantification via applying the computerized simulation tool The Dortmunder. For the complete posture replication of the observed person, the data of the angular position of each body segment were reproduced on the basis of the video recordings and a manikin was presented using a stick figure animation. With specific software, the 3D coordinates of each joint and body segment were determined in a body-related coordinate system. In a recursive procedure, the coordinates were changed and the position and movement of the stick figure were adapted to the position and movement of the infrared markers (Jordan et al., 2003).

**Determination of action forces**

With the help of newly developed devices—a measuring bed, different chairs, bathtub, and floor—the caregiver’s action forces applied to the patient during a transfer were captured continuously during the transfer. The forces were not measured ‘directly’ at the hands of the caregiver, but ‘indirectly’ by the change of the forces transferred to the measuring device, i.e. the change in the reaction forces of the measuring bed etc., is considered to figure the action forces applied by the nurse. The forces had to be captured regarding amplitude, direction, and the point of application. When configuring the measuring systems, special emphasis was put on copying the structure and the functionality of the underlying devices as exactly as possible (Theilmeier et al., 2003, 2006).

For the construction of the measuring bed (see Fig. 2, left-hand part), a common hospital bed was equipped with an additional framework attached to the bedstead and connected to the bedspring frame via triaxial piezo-ceramic force sensors at the four bed corners. To prevent the bending of the bedspring frame, which results in malfunction of the force sensors, the bed was stabilized by inserting an additional frame structures and several supporting stands. Additionally, two ‘measuring bars’, each equipped with two triaxial force sensors, were mounted on the bed—one at the bed’s long side, the other at the bed’s head—allowing the measurement of forces induced by a leaning of the HCW against the bed. Two force platforms at the floor were used to capture the ground reaction forces of the HCW when the patient leaves the bed and so
the measuring function of the bed fails. To enhance the contact of the force platforms to the laboratory floor and to level surface unevenness, the platforms were placed on concrete plates.

The basis for the determination of the action forces is described representatively for all used force measuring devices in the following for the measuring bed: The action–force amplitude and its direction, i.e. the ‘overall resultant’ action force result from the vectorial sum of all force components measured with the triaxial force sensors at the four bed corners—diminished by the force due to the patient’s weight. For the calculation of the point of force application (distance between action force vector and the reference point at the middle of the bed’s surface), the vectorial sum of moments could be drawn with the applied moment, the measured moment, and the moment caused by the patient’s weight (‘patient’s moment’). The applied moment contains the overall resultant measured action force and the associated point of force application that is searched. To determine the patient’s moment at each point in time during the transfer, the patient’s centre of gravity shift was gathered with the help of accompanying biomechanical model calculations considering patient’s movement during the transfer.

The combination of the measuring bed, the measuring bar at the bed’s long side, and the force platforms, furthermore allow the calculation of the force distribution when two HCWs handle a patient at the same time. The action force of the first caregiver can be calculated from the ground reaction forces diminished by the forces measured with the bar in due consideration of the respective point of force application. The action force of the second HCW is determined from the force measured with the bed diminished by the action force of the first HCW.

To examine transfer activities to a chair or from a chair, the system ‘measuring chair’ was developed on the basis of a commonly used patient chair (see Fig. 2, right-hand part, above). In a first version of the measuring chair, the patient chair was mounted on a single force platform with four 3D force sensors; it was used to gather the action forces of the HCW during simple transfer activities like ‘raising a patient from sitting to upright standing position’. For more complex patient movements, the system was modified applying two additional force platforms (see Fig. 2, right-hand part, above). Determining action forces with the system, measuring chair works according to the same principle as described above for the measuring bed.

Due to the dimensions and characteristics of the patient chair (e.g. existence of lowerable arm rests or material of the upholstered seat), it was possible to reproduce many actions commonly performed in the care sector like ‘placing a patient from sitting at bed’s edge in a chair and vice versa’ or ‘raising a patient from sitting to upright standing position and vice versa’ with high authenticity. The height of the measuring chair can be adapted according to the respective requirements. Since the forces transferred by the HCW to the patient should be measured, a contact of the caregiver to the force platform was avoided by using two footstep bridges positioned above the platform.

For the determination of the nurse’s action forces during the activity ‘moving the patient into the
bathtub’, the system ‘measuring bathtub’ (see Fig. 2, right-hand part, below) was developed. In order to simulate the care situation close to reality, the dimensions and geometry of the measuring bathtub were arranged according to a commonly used bathtub. For this purpose, a usual bathtub was installed onto a steel framework. Forces, applied on this construction, were measured with the help of two force platforms between the framework and the laboratory floor. The contact of the framework to the force measuring platforms was established using four contact elements adjustable in height in order to guarantee a uniform force transmission. The height of the standing area of the caregiver could be adjusted using pedestals of various heights.

For the analysis of patient transfers on the floor, like ‘raising a patient from sitting on the floor to sitting or to upright standing position’, the nurse’s action forces were measured using the measuring system ‘floor’. With the help of two force platforms, arranged side by side, the HCWs’ action forces were captured indirectly by the measurement of the changes of the ground reaction forces. The adaptation of the standing area of the HCW to the overall height of the force platforms and concrete plates was obtained by using several pedestals. Since the friction between the patient and the metallic surface of the force platform was too small in comparison to real conditions, so-called anti-slide mats were positioned on the force platforms. By this measure, an effective support of the HCW was achieved and close to reality conditions were established.

The lumbar load of the HCW was determined by model calculations applying the 3D dynamic simulation software tool The Dortmunder. This tool permits the quantification of several low back load indicators considering the gravitational and inertial effects of the body and a handled object or—here—a subject, the effects of asymmetry of posture, movement, and force exertion, as well as the effects of intra-abdominal pressure in supporting the trunk during forward-inclined positions. The lumbar load is described by The Dortmunder on the basis of mechanical characteristics, like compressive or shear forces and bending or torsional moments, with respect to the intervertebral discs in the lumbar section between the first lumbar vertebra (L1) and the upper part of the sacrum (S1). The calculations in this paper concern the lowest intervertebral disc in the spine called ‘L5–S1’ as reference point. The lower lumbar spine is generally accepted as a ‘bottleneck’ since it represents the structure with very high risk for the development of lumbar degenerative diseases.

**TYPICAL APPLICATION**

The measuring systems were developed in order to study the lumbar load of HCWs during manual patient handling actions, which are presumably accompanied with high biomechanical load on the lumbar spine. In the following, exemplary measurements and calculations regarding the activity ‘lifting a leg of a lying patient and vice versa’ will be demonstrated. In Fig. 3, the posture of the nursing person is represented at three basic points in time (postures B, C, and E as described in the following) on the basis of three photos.

To obtain input data for the biomechanical model calculations, the posture during the transfer activity was reproduced with the help of the optoelectronic data and by video recordings as described in Posture recording.

The activity was divided into seven segments (marked with the characters A–F) because with the biomechanical model only unidirectional segment movements can be analysed within one trial. The segments are represented in Fig. 4 as stick figures in side view: nurse standing in an upright position (A), bending the trunk (A–B), lifting the patient’s leg (B–C), holding the leg (C–D), lowering the leg (D–E), nurse straightening up (E–F), and standing in upright position again (F). The activity was partly performed by the HCW with clearly inclined trunk (positions B and E in Fig. 3). In addition, for the execution of the activity, the nurse adopted a slightly side-bent posture (positions B and E in Fig. 3).

Represented in Fig. 5, time courses of the horizontal (forward/backward and leftward/rightward) and vertical (upward/downward) action force components were recorded for the activity lifting a leg of a lying patient or vice versa. The segments are marked with the same characters as the stick figures.

In the sections ‘HCW upright’, the HCW was in the so-called basic position without applying any action forces. In the second section, the caregiver bent down to the patient and—in order to grasp her leg to raise—supports herself on the bed. Resulting from that at the end of the section, a force of 70 N upward and up to 50 N in vertical direction (downward) was determined. The third section ‘lifting patient’s leg’ is characterized by a steep raising of the upward vertical force component up to 90 N and a horizontal lateral force component changing from 70 N downward to 30 N leftward. During the section ‘holding patient’s leg’, the caregiver held her leg using an upward force of 75 N and exercised an easy force backward to the patient. In the next section ‘lowering patient’s leg’, the vertical force component declines.
to zero when the leg lays on the bed surface and becomes negative, as soon as the caregiver supports herself on the bed. The activity is finalized by the sections ‘straightening up’ and ‘HCW upright’ with action forces similar to the corresponding sections at the beginning of the activity.
With the help of posture and force data, as indicators for the lumbar load, time courses of forces and moments at the lumbosacral disc (L5–S1) were calculated by applying the biomechanical model. The time courses of the force components at the lumbar disc (shear force forward/backward, shear force leftward/right, and the axial compressive force) are presented in Fig. 6.

In the sections called HCW upright (posture A), disc forces were determined, which are typical for an upright standing person. The compressive force amounted to ≈0.7 kN due to the weight of the caregiver's body segments superior to L5–S1, the sagittal 'forward' shear force adds up to 0.2 kN due to anatomically caused tilt of the disc, a lateral 'sideward' shear force is missing because of the symmetrical posture of the caregiver in this phase. In the second section (the HCW bends forward to the patient, motion from posture A to posture B) due to only small action forces applied by the caregiver, the typical influence of a motion with acceleration and deceleration shows up: the compressive force (0.7 kN) is approximately as high as the value for the basic posture at the beginning of the section, is slightly increased to ≈1 kN to overcome the resting status, rises then up to a maximum of ≈2.5 kN (maximum deceleration when retracting the trunk in motion), and drops finally to ≈2 kN (decreased deceleration of the trunk to receive the intended inclined posture). The compressive force achieved a maximum of 4 kN—the highest value for this activity—at the beginning of the third section (postures B–C) because of a clearly forward bent posture of the caregiver on the one hand and because of the acceleration of both, the own moved body segments and the mass of the patient's leg on the other hand. During holding patient's leg (postures C–D), the disc compression remains in a range of 3 kN. The two sections lowering patient's leg and straighten up essentially correspond reversely to the process in the sections 'lifting' and 'bending'. The time courses of the shear forces follow the time course of the compressive force approximately—however, with clearly smaller values (highest values: 0.6 kN sagittal and 0.5 kN lateral).

The time courses of the shear forces follow the time course of the compressive force approximately—however, with clearly smaller values (highest values: 0.6 kN sagittal and 0.5 kN lateral). The activity described exemplarily above lifting or lowering a patient's leg was performed by two persons acting alternately as a HCW or as patient (P1 and P2). The time courses of lumbar-load indicators were analysed for totally nine executions were analysed (five and four times for Patient 1 or 2, respectively). Regarding the peak value in a time course of the indicator 'compressive force at L5–S1', the average amounts to 3.3 kN for P1 (range: 2.9–4.0 kN, n = 5) and 2.9 kN for P2 (2.6–3.5 kN, n = 4).

**DISCUSSION**

**Criticism of the methods**

The methodology described in this paper is applied in a research project concerning the determination of indicators for the lumbar load of HCWs during patient transfer activities. For this purpose, measuring systems were implemented to determine the main influencing factors on lumbar load, such as the posture of the HCW and the forces transferred to the patient. Prior to the laboratory measurements simulating typical patient transfer activities in

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**Fig. 6.** Time courses of the resulting lumbar load, i.e. of the forces transferred via the lowest intervertebral disc of the spine ('L5–S1'), for the care activity ‘lifting a leg of a lying patient and vice versa’ in conventional execution with a passive patient.
In particular, the linearity, repetition accuracy, crosstalk, and the temporal drift for the entire force measurement systems consisting of the bed, chair etc., with the integrated force sensors or force platforms were determined. In the following, some characteristic findings are shown exemplarily, in particular for the measuring bed. More detailed information and results of the other measuring systems are provided by Theilmeier (2006).

In several series, the force measurement systems were charged with forces of different amplitude and different directions. For the coefficient of determination ($R^2$), a value of $>0.99$ was found in each case indicating the high linearity of the force sensors, even if they are inserted in the measuring systems. By the use of a calibration in the relevant measuring range, the absolute values for measurement errors were $<25$ N in vertical direction and $<20$ N in horizontal direction.

The reliability of the systems was examined on the basis of in total 10 measurements for each direction. A maximum deviation between the readouts of $<5\%$ was always observed. As to the crosstalk between the signals representing the three spatial force components, values of $<3\%$ were measured; for example, loading the measuring bed with a vertical force of 850 N results in crosstalk signals in the horizontal directions of 25 N ($\approx 3\%$, transverse bed axis) and of 10 N ($\approx 1\%$, longitudinal axis). Because the used piezo-ceramic sensors exhibit an unavoidable electrical drift, the temporal change in the readouts was examined. The measured drift corresponds to force changes of $0.3$ N min$^{-1}$ (if loaded with 270 N vertically) and $0.5$ N min$^{-1}$ (unloaded). Since the duration of the examined patient transfers is clearly $<1$ min, the ‘drift influence’ can be regarded as negligible.

A force applied to the measuring bed results in reaction forces at the four edges of the bed where the force sensors are located; the readouts of the four sensors depends directly on the localization of the point of force application (PoF). Therefore, inaccuracies with the measurement of the forces, described above, have a direct influence on the accuracy of the localization of the point of force application whose determination results from fixed geometrical dimensions of the measuring bed. Therefore, the verification of the point of force application accuracy is studied exemplarily in detail. The bed’s surface is shown schematically in Fig. 7 together with the insertion of 11 measuring grid points (closed circles) where a vertical force of 150 N was applied successively in five series of measurement. An additional weight of 68 kg, continuously placed close to the

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_Table 1. Results of the inspection of measuring accuracy-values in millimeter; set point $= 202$ mm_

<table>
<thead>
<tr>
<th>$n$ = 100</th>
<th>Measuring condition</th>
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<th>45°</th>
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<tbody>
<tr>
<td>Mean value (mm)</td>
<td>202</td>
<td>202</td>
<td></td>
</tr>
<tr>
<td>Standard deviation (mm)</td>
<td>$&lt;1$</td>
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a hospital, quality characteristics of the measuring systems were checked.

_Posture recording._ The optoelectronic posture measuring system is equipped with two position sensors (see Fig. 1); each of them is assembled with three infrared cameras. The accuracy of the position detection of the infrared markers used to track the location and the movement of the nurse’s body segments depends on the adjustment of the cameras within the position sensors. This adjustment was already performed by the manufacturer. Nevertheless, after fixing the position sensors at the walls of the laboratory, the overall accuracy of the total device was checked.

For this purpose, two infrared markers were fixed on a solid bar in a definite distance of $\approx 20$ cm. The bar was positioned at different places within the measuring space of the laboratory (see Fig. 1) and the distance between the markers was determined using the total measuring device. The measurement was performed using two different orientations of the bar, one with the bar in parallel to the position sensors and another with an angle of 45°. The results of 100 measurements are shown in Table 1: the optoelectronically measured intermarker-distance corresponds to the ‘real’ distance measured with a sliding rule and the standard deviation is $<1$ mm, i.e. $<1\%$, for both orientations of the bar.

The examination of the measuring accuracy substantiated that the used measuring system itself operates with a very high accuracy. For the measurement of the position of HCW’s body segments, the overall accuracy essentially results from additional factors, such as the positioning of the markers distant from the real body joints on the subjects skin or clothing. Some disadvantages of an optoelectronic system usage—in particular, the occasional covering of the markers by body segments of the HCW or the patient—was compensated for by using the video systems’ data. In such cases, the combination of the optoelectronically measured data with the video images allows to complete the position data for relevant periods of time.

_Determination of action forces._ For the newly developed force measuring systems, comprehensive testing and calibration procedures were performed.
bed’s centre, simulated the patient (rhombus in Fig. 7). As results of these measurements and the respective computations, the calculated positions of the points of force application are shown as open circles. The deviation from target value (closed circle) to actual value (open circle) amounted to \( \pm 40 \) mm (mean value 37 mm, minimum 15 mm, maximum 91 mm, standard deviation 16 mm). Related to the measuring range (2000 mm), the average deviation is \( <2\% \), the maximum deviation was \( <5\% \).

The dynamic behaviour of the measuring device is only of subordinated importance: For example, the natural frequency of the measuring bed clearly exceeds 20 Hz, whereas a total patient handling lasts few seconds with activity sections of about half a second. Nevertheless, the performed activities are to be considered highly dynamic from the ergonomic point of view; the influence of accelerations of patient’s and caregiver’s body masses including inertial effects has therefore to be considered in lumbar-load quantification, which is guaranteed, here, by applying the multi-linked 3-D dynamic validated simulation tool The Dortmunder.

Comparison with recent studies

The aim of the research project described here was the estimation of lumbar load during care activities on basis of measurement-supported posture and action force determination considering spatial and inertial influences. In spite of their doubtless merits with other aspects in scientific analysis and prevention, previous biomechanically oriented studies into nurse’s low back load can serve as a reference under restriction only (Waters et al., 1993; Marras et al., 1999; Elford et al., 2000; Soyka, 2000; Skotte et al., 2002; Schibye et al., 2003; Garg, 2006). In these studies, for example, other types of patient transfers were analyzed or the transfers were limited to ‘best-case scenarios’ with low-weight patients or advantageous posture of the nursing person. In other cases, inertial effects due to patient-and-HCW movements or horizontal action force components were disregarded.

A systematic overview particularly to studies with emphasis placed on ‘patient transfer’ was provided by Hignett (2003) and Hignett and Crumpton (2007) and Hignett et al. (2004). The authors consider both different transfer techniques and intervention possibilities, which are able to reduce the load for the nursing stuff. In this context, Freitag et al. (2007) particularly analysed the loads of HCWs caused by posture during whole shifts. In contrast to the procedure of quantitative determination of short-term mechanical load—aimed to the assessment of an acute overloading risk for lumbar-spine elements, presented here—‘dose-oriented’ procedures are well suited for the description of the cumulative load over the working day or the entire working life by considering the correlation of loading duration, frequency, and intention, in particular. The assessment of such dose values take into ancient criteria based on epidemiological studies about the correlation of long-term mechanical load and the development of lumbar degenerative diseases (Seidler et al., 2009).

Conclusions

In the presented study, the lumbar load of HCWs during manual patient handling is described mostly
on basis of the force and their components at the disc L5–S1. The selected ‘3D procedure’ also makes statements possible about the load aspects regarding bending and torsional components of force for the lower spine, which are acting simultaneously as the forces and require special attention regarding the development of overload-induced disease. In conclusion, aiming an appropriate indication of lumbar load during patient transfer, the described example demonstrates the necessity of a time-variant posture-and-force determination and a sophisticated biomechanical model. This methodology, however, is accompanied by a high level of personnel effort for data gathering and evaluation, and therefore, it is less well suited for larger subject groups. A detailed analysis of the subject’s task execution with the help of the posture-and-force data and a careful selection of typical tasks from the pool of the totally collected data could reduce the disadvantages of having a low sample size as in this laboratory study.

**Forecast**

As the analyses of ~160 representative transfer actions studied in ‘the Third Dortmund Lumbar Load Study’ have shown patient transfer activities may result in intensive lumbar load for the HCWs, which in many cases exceed recommended lumbar-load limits, such as NIOSH’s (1981) action limit or the age-and-gender specific ‘Dortmund Recommendations’ (Jäger et al., 2001b) for the assessment of manual handling activities with respect to potential biomechanical overload of the spinal structures in the lumbar region. Work evaluation with the obtained data will allow the derivation of preventive work design measures: the testing of optimized transfer techniques and of small aids (e.g. sliding mat or glide board in order to lower the friction between the patient and the bed-or-chair surface) will presumably show that lumbar load can furthermore be lowered in order to achieve health prevention in the everyday working life of healthcare workers.

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**REFERENCES**


