Lumbo-sacral loads and selected muscle activity while turning patients in bed

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Handling patients in bed using a piqué (a waterproof padded sheet placed under the patient) is associated with a high incidence of risks for the spine with, in particular, the activity of pulling and turning the patient with the piqué representing the highest risk. Fifteen female nursing aides were evaluated for compression and shear forces at the L5/S1 joint and for selected muscular activities in the trunk and shoulders. Films, force platforms and EMG recordings supplied the data; dynamic segmental analyses were performed to calculate reaction forces at L5/S1, and a planar single-muscle equivalent was used to estimate internal loads. Different execution parameters were examined including execution velocity, height of bed, direction of effort, leg position and knee support. A ‘free’ task, and a manual task not involving the piqué, were also investigated. Recommendations are made for reducing spinal loading. The results also suggest that a change of direction in the trunk motion may present some risks when associated with handling of heavy loads. Furthermore transfer of problems from a particular joint to other joints is likely to occur.

1. Introduction

On the basis of an extensive survey of the literature, Andersson (1981) highlighted the frequency and severity of lower-back injuries among workers in industrialized countries, as well as the high cost of these injuries to society. Nursing aides, a group of workers who provide basic and non-medical patient care, are particularly liable to lower-back injuries (Delhin et al. 1976). Among the potential risk factors, patient handling is frequently cited, but the specific tasks which might be more strenuous and hazardous to the spine have rarely been examined. Furthermore, current safety training programmes generally appear to be based on empirical observation and common sense rather than on carefully controlled scientific study.

Imbeau and Lortie (1984) analysed occupational accidents affecting nursing aides employed in a geriatric hospital. Back injuries were the most frequent results, with an incidence of 62%, and injury to the lower-back region accounted for two-thirds of these accidents. The incidence of accidents to the upper arms is also particularly high among female nursing aides and represents 22% of all accidents.

A subsequent ergonomic study by Lortie (1985) found that the activities presenting the greatest risk were primarily associated with horizontal effort like turning the patient over in bed. This was the most frequent operation required of female nursing aides and was associated with 19% of all accidents. This task consists of two phases: translation of the patient over to the side of the bed followed by rotation, after which the patient is lying on his or her side in the centre of the bed. When handling patients in bed, the female nursing aides in the hospital studied generally used the piqué as an aid, the piqué being a waterproof and padded sheet placed under the patient. The piqué was
implicated in 16% of all accidents involving female nursing aides. The act of turning a patient over in bed using a piqué was therefore selected for thorough biomechanical analysis.

The biomechanical approach in the present study is based on estimation of spinal loads occurring during the movement with special emphasis on peak values. These estimates provide important input for the development of preventive measures. This approach led the National Institute for Occupational Safety and Health (NIOSH, 1981) to propose maximal allowable limits: compressive loads not exceeding 3430 N require no assistance; action is required for heavier loads with a maximum allowable limit of 6377 N. Evaluation of the loads on the spine is usually done using mathematical models (Chaffin 1969, Gracovetsky et al. 1977). Until recently, most of these models were quasi-static, with the acceleration effects being assumed to be negligible. Recent studies have shown however that these effects are appreciable (Leskinen et al. 1983, Freivalds et al. 1984, McGill and Norman 1985). Because of the nature of the motion for analysis in the present study, it was felt necessary to develop a dynamic segmental model. The task analysed involves forces being exerted horizontally against static and dynamic friction forces between the piqué and the bed-sheet and shear loads on the spine might be significant.

Analysis of the loads that the lumbo-sacral joint might be subjected to in the course of work activities cannot be considered complete without a concurrent study of other joints. It is possible that performance of a given task may reduce loads on the spine but by the same token increase loads on other joints such as the shoulders and knees: in that case, one would simply be witnessing a transfer in the problem region. Due to the incidence of accidents to the upper arms, this particular problem was partially explored in this investigation, and restricted to EMG recordings of the deltoid muscles, supplementing EMG data on the erectors, rectus abdominis and obliques.

Based on field observation data, the task consisting of translating and turning a patient in bed using a piqué was analysed for several performance parameters: execution velocity, height of the bed, direction of effort, leg position and support against the bedside. A ‘free’ task, i.e., as done the way each nursing aide normally performed the task, was also analysed, as was, finally, a manual task, which excluded use of the piqué. The purpose of the study was to examine the above parameters for their effects on the kinematics and kinetics of the motion. Evaluation of spinal loads at L5/S1 and EMG activity for selected muscle groups provides important information for the development of preventive measures and the institution of safe training methods.

2. Methods

This section discusses the development of a dynamic model for the estimation of loads at L5/S1 and consists of the kinematic segmental analysis, dynamic segmental analysis and modelling of the loads at L5/S1. The experimental procedures will be described at the end of the section and will include subjects, tasks, measurement techniques and statistical analyses.

2.1. Kinematic segmental analysis

Subject’s bodies were subdivided into 11 segments which were assumed to be rigid: both feet, legs, thighs, lower trunk (below L5/S1), upper trunk, head, plus both upper arms and both forearms and hands, which were treated as single segments since they work symmetrically. Body segment parameters were determined from the data
published by Dempster (1955) for segment masses and location of centres of gravity; the segment moments of inertia about a transverse axis were obtained from Zatsiorsky and Seluyanov (1983).

The anatomical coordinates of the joint’s centres were smoothed and differentiated with quintic splines using a least squares approximation (Dierckx 1975) to yield velocity and acceleration data. The estimates of velocity and acceleration are generally more imprecise at both ends of the data sets. To improve estimates, five extra points were digitized at the beginning and end of the movement. These procedures have been advocated in the literature (Wold 1974, McLaughlin et al. 1977). Our choice of five points was based on the analysis of selected trials with 5, 10 and 15 points at the extremities where the results showed the absence of significant differences in the dependent variables related to the number of points.

The coordinates of the segments’ centres of gravity as well as their horizontal and vertical velocities and accelerations could be determined for any film sequence from the linear relationships applicable to rigid bodies. For angular velocities and accelerations, the principles of relative motion of a rigid body were utilized to describe the motion of a link joining the two segment end points.

2.2. Dynamic segmental analysis

For calculations of net reaction forces and net muscle moments at each joint, a dynamic analysis was performed on each body segment. The segmental model included the net reaction forces and moments at each joint, the external forces either measured by force platforms (feet, knee) or solved for as unknowns (hand forces), the segment’s weight, with the inertial forces and moments ($ma$ and $I\alpha$) being determined from the kinematic analysis. The analysis covered each segment, starting at the feet and working up adjacent segments successively until the L5/S1 level was reached. The analysis was also extended to estimate the external forces exerted on the patient at hand level.

Figure 1 gives a diagram of all external forces, known and unknown, acting on the subject. Figure 2 presents the dynamic segmental approach for the foot segment linked to the leg, where the following equations of dynamic equilibrium were used:

\[ \Sigma F_x = F_{x_i} + R_{x_i} = m_i(a_{x_i})_G \]

\[ \Sigma F_y = F_{y_i} + R_{y_i} - W = m_i(a_{y_i})_G \]

\[ \Sigma M_o = M_i + (F_{x_i})(d_{x_i}) + (F_{y_i})(d_{y_i}) - (W - (L \times G)) \cos \theta = I_o \alpha = \theta \]

The only unknown values are for the reaction forces $R_{x_i}$ and $R_{y_i}$ and the muscle moment $M_i$, which are solved for and subsequently served as input data for segment analysis of the leg. Subscript $i$ is the segment identification, $G$ is for centre of gravity, $d$ for lever arm, $W$ for weight, $L$ for segment distance and $F$ for external force.

2.3. Modelling the loads at L5/S1

The lower part of the body is presented in figure 3 with free-body diagrams for a cross-section at L5/S1. The model of the spine was developed to estimate compression and shear forces as well as the single-equivalent musculo-ligamentous force which provides the extensor moment of the back (Gagnon et al. 1985, 1986). The intra-abdominal force was ignored in the model for the present study since its role in loading the spine is the subject of considerable controversy in the literature (Gilbertson et al. 1983, Gracovetsky et al. 1981).
From the net reaction forces and moment found at L5/S1, the internal forces were estimated from the equations of equilibrium:

\[ R_c = F_c - E \]  
\[ R_s = F_s \]  
\[ M_o = Ea \]

2.4. Subjects

Fifteen female nursing aides volunteered to participate in this study; their mean age was 34.5 years (range: 19–52 years), their mass, 62.4 kg (46–84 kg), their height, 163.5 cm (154–173 cm) and their degree of experience, 6.6 years (1–15 years). All subjects were familiar with the task involving use of a piqué. The role of the patient was assumed by a healthy female subject weighing 52.0 kg and 158.7 cm tall.

2.5. Tasks

Each subject did two repetitions of six different variations in task execution which allowed the following comparisons: direction of force (horizontal and vertical pulls: tasks 3 and 4), execution velocity (slow and rapid: tasks 4 and 6), leg position (symmetric vs. asymmetric: tasks 7 and 3), the knee support on the bedside (present or absent: tasks 3 and 5), the height of the bed (low and high, here referring to vertical distances of 16 and 6 cm between the plane of the bed and the forearm when flexed at 90°: tasks 4 and 8). For all main effects tested, the other factors were all controlled for. The treatment conditions also included a ‘free’ task (task 1), defined as use of the piqué as the subject normally used it during her routine activities. Finally, there was a manual
task (task 2), where the piqué was not utilized and which involved bracing the arms around the patient's shoulder and thigh. These latter two tasks were not controlled for any factor. The tasks are illustrated in figure 4. Tasks 6 and 1 were not shown on the figure: task 6 was similar to task 4 for the body positions represented since only the speed of execution distinguished both tasks; task 1 varied for each subject. A total of 8 treatments and 16 trials were performed by each subject.

2.6. Measurement techniques

The measurement techniques used were three-fold: cinematography, force platforms and electromyography. A 16 mm Locam camera set at a speed of 50 frame/s was used to record the subject's body positions during the movement. The films were analysed by a NAC motion analyser including a graf/pen digitizer and linked to a PDP-11/23 mini-computer.

One AMTI and two Kistler force platforms were used to record the external forces and their point of application on both feet and the knee(s). The horizontal and vertical components of the forces in the main plane of motion were included.
EMG activities were monitored with surface electrodes for three muscle groups: the erector spinae on the subject's left side and the medial portion of the deltoids on both shoulders. For three subjects, activity in the rectus abdominis and the obliquus externus on the left side was also monitored. The position of the electrodes on the deltoids was based on recommendations by Basmajian et al. (1980) and the positions for the other muscles were based on recommendations by Andersson et al. (1977) and Ekholm et al. (1982). The EMG signals were amplified and filtered with a 15–1000 Hz pass band and recorded on FM tape; the PDP-11/23 mini-computer was used to digitize the data with a sampling rate of 1000 Hz. A spectral analysis of the EMG signals using a Fast Fourier Transform (FFT) was done to verify if aliasing errors could be introduced in the data. This analysis revealed that there was not a significant content of noise above 400 Hz. For the type of analysis based on periods of integration of 20 or 250 ms, the changes which might have been attributed to aliasing errors were not apparent.

EMG activity was expressed as a percentage of activity in the particular muscle during maximal voluntary isometric contraction. The signals were rectified and integrated for periods of 20 ms; for maximal contractions, the integration period was extended to 250 ms. To yield the relative percentages, the data were adjusted to the same time basis. The curves thus obtained were smoothed by means of a second-order moving average filter. The positions for maximal isometric contractions are described in Ekholm et al. (1982) for the rectus abdominis and obliquus externus muscles and in Grieve and Pheasant (1976) for the erector spinae. For each deltoid, the subject

Figure 3. The model. (A) Net reactions and (B) internal forces at the lumbo-sacral joint including compression force $F_c$, shear force $F_s$ and muscular force for the erectors $E$. 
assumed a standing position with the arm at 180°, completely extended above the head, with the palm of the hand turned outward; in this position, the subject had to resist an external force abducting her arm.

Synchronization of film sequences, force and EMG data was accomplished by means of an electrical pulse or step increase in voltage recorded on a specific channel reserved for synchronization and generated both on the computer and the FM tape. Simultaneously, a light emitting diode (LED) appeared in the foreground of the film frames.

A questionnaire was also administered on the nursing aides’ perceptions regarding the degree of difficulty of each task. Based on a 5-points scale, it describes perceptions as very easy, easy, moderate, difficult and very difficult.

2.7. Statistical analyses

Analyses of variance involving repeated measures were first conducted on the two repetitions of the eight treatments. No significant differences were found between the two repetitions. Therefore, the data obtained on each dependent variable were averaged for each subject and each treatment, and the analyses of variance for repeated measures were subsequently conducted on the treatments only.
For those variables where significant differences were obtained, univariate or Scheffé tests were applied to locate the differences. A 0.1 level of probability was chosen. Univariate tests were used for planned comparisons including direction of force (tasks 3 vs. 4), execution velocity (tasks 4 vs. 6), knee support (tasks 3 vs. 5), leg position (tasks 3 vs. 7) and height of the bed (tasks 4 vs. 8). Scheffé tests were used to compare the free task (task 1) and the manual task (task 2) both with each other and with all other tasks.

3. Results

This section describes the general results relative to the kinematics of the task, task kinetics including external forces and spinal loads and EMG data. The different parameters of task execution are also compared.

3.1. General results of kinematics, kinetics and EMG

In this section, all results are presented in relation to the kinematics of task execution, the kinetic variable and EMG data.

3.1.1. Kinematics of the task: These results are presented in table 1. They show that the various modes of execution differed substantially and that there was high between-individual variability in the dependent variables analysed. In general, the movements associated with the successive actions of pulling and turning the patient tend to be of relatively long duration, between 1.4 and 2.0 s. The pulling and turning phases lasted, respectively, about one-third and two-thirds of the total movement time; the transition occurred when the hip joint changes direction from backwards to forwards. The amplitude of movement as reflected in the displacement of a marker at L5/S1 during the pulling phase varied between 6 and 24 cm, depending on the main direction of effort. The trunk was also moderately inclined at the start, except for the manual task where the trunk was observed to be near horizontal.

3.1.2. Kinetics: external forces and spinal loads: The kinetic results (table 2) include the external forces the nursing aide applies to the patient and the internal forces represented by the maximal shear and compression forces at L5/S1 and the effect over time of spinal compression loading.

The mean values for maximal compression forces ranged between 2479 and 3526 N, a 30% difference, whereas the spread in the maximal shear forces was only 554 to 661 N, a 16% difference. Shear forces were considerably lower than compression forces and usually varied very little during the movement. The maximal compression forces generally were found to occur at or near the transition point. These internal force patterns are presented in figure 5, for a typical case.

The patterns of external forces exerted on the patient are illustrated for two typical cases, one horizontal and one vertical task (figure 6). The vertical task was characterized by a very substantial reduction in horizontal forces and only a slight increase in vertical forces. This was observed in all subjects. The results showed that very little in the way of horizontal forces is exerted during the turning phase, where the horizontal forces should change from positive to negative. Observation of the movement showed that when the hips initiated their change of direction from backwards to forwards, the arms were seen to enter into vigorous action, being drawn backwards in hyperflexion. The highest forces were therefore observed near the transition point.
Table 1. Kinematic factors for the different tasks (N = 15).

<table>
<thead>
<tr>
<th>Variables</th>
<th>Tasks</th>
<th>1* Free task</th>
<th>2* Manual task</th>
<th>3* $F_H \cdot V_R \cdot B_L \cdot K_{S-L_A}$</th>
<th>4* $F_V \cdot V_R \cdot B_L \cdot K_{S-L_A}$</th>
<th>5* $F_H \cdot V_R \cdot B_L \cdot K_{N-L_A}$</th>
<th>6* $F_V \cdot V_S \cdot B_L \cdot K_{S-L_A}$</th>
<th>7* $F_H \cdot V_R \cdot B_L \cdot K_{S-L_T}$</th>
<th>8* $F_V \cdot V_R \cdot B_H \cdot K_{S-L_A}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Duration of motion(s)*</td>
<td>X</td>
<td>2.1</td>
<td>1.9</td>
<td>1.7</td>
<td>1.4</td>
<td>2.0</td>
<td>2.0</td>
<td>1.7</td>
<td>1.5</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td>0.4</td>
<td>0.4</td>
<td>0.3</td>
<td>0.3</td>
<td>0.4</td>
<td>0.4</td>
<td>0.3</td>
<td>0.3</td>
</tr>
<tr>
<td>Time of transition/total duration (%)*</td>
<td>X</td>
<td>29.0</td>
<td>35.9</td>
<td>34.0</td>
<td>34.5</td>
<td>36.5</td>
<td>33.4</td>
<td>31.9</td>
<td>28.7</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td>10.0</td>
<td>13.0</td>
<td>9.4</td>
<td>15.3</td>
<td>9.3</td>
<td>14.6</td>
<td>11.0</td>
<td>12.9</td>
</tr>
<tr>
<td>Average vel. at the hands (m/s)*</td>
<td>X</td>
<td>0.5</td>
<td>0.3</td>
<td>0.6</td>
<td>0.6</td>
<td>0.5</td>
<td>0.4</td>
<td>0.6</td>
<td>0.6</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td>0.1</td>
<td>0.1</td>
<td>0.2</td>
<td>0.2</td>
<td>0.1</td>
<td>0.1</td>
<td>0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>Initial acc. at the hands (m/s^2)*</td>
<td>X</td>
<td>2.3</td>
<td>2.6</td>
<td>2.6</td>
<td>2.5</td>
<td>3.4</td>
<td>2.4</td>
<td>2.6</td>
<td>2.4</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td>1.3</td>
<td>1.7</td>
<td>1.5</td>
<td>1.4</td>
<td>1.8</td>
<td>1.2</td>
<td>1.3</td>
<td>1.6</td>
</tr>
<tr>
<td>Average acc. at the hands (m/s^2)*</td>
<td>X</td>
<td>2.1</td>
<td>1.5</td>
<td>2.9</td>
<td>3.6</td>
<td>2.6</td>
<td>1.5</td>
<td>2.9</td>
<td>3.1</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td>0.8</td>
<td>0.4</td>
<td>1.0</td>
<td>1.5</td>
<td>1.1</td>
<td>0.5</td>
<td>0.9</td>
<td>1.0</td>
</tr>
<tr>
<td>Hor. disp. of L5/S1 total (cm)*</td>
<td>X</td>
<td>14.5</td>
<td>14.8</td>
<td>16.6</td>
<td>7.2</td>
<td>24.4</td>
<td>5.9</td>
<td>16.1</td>
<td>4.5</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td>9.1</td>
<td>9.0</td>
<td>7.7</td>
<td>4.3</td>
<td>9.9</td>
<td>4.8</td>
<td>8.1</td>
<td>3.4</td>
</tr>
<tr>
<td>Initial back angle with vert. (°)*</td>
<td>X</td>
<td>27.3</td>
<td>67.8</td>
<td>35.6</td>
<td>37.6</td>
<td>40.8</td>
<td>37.1</td>
<td>34.4</td>
<td>29.8</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td>8.8</td>
<td>5.3</td>
<td>9.2</td>
<td>9.5</td>
<td>9.5</td>
<td>9.9</td>
<td>7.6</td>
<td>10.0</td>
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</table>


Sign. diff.: 3–4; 4–6; 3–5; 1–3; 1–4; 1–7; 1–8; 2–4; 2–8.

Sign. diff.: 4–8.

Sign. diff.: 4–6; 3–5; 1–2; 1–3; 1–4; 1–6; 1–7; 2–3; 2–4; 2–5; 2–6; 2–7; 2–8.

Sign. diff.: 3–5.

Sign. diff.: 4–6; 3–5; 1–2; 1–3; 1–4; 1–6; 1–7; 1–8; 2–3; 2–4; 2–5; 2–7; 2–8.

Sign. diff.: 3–4; 3–5; 1–4; 1–5; 1–6; 1–8; 2–4; 2–5; 2–6; 2–8.

Sign. diff.: 4–8; 3–5; 1–2; 1–3; 1–4; 1–5; 1–6; 1–7; 2–3; 2–4; 2–5; 2–6; 2–7; 2–8.
Table 2. Kinetic factors for the different tasks including external forces and spinal loads at the L5/S1 joint (N = 15).

<table>
<thead>
<tr>
<th>Variables</th>
<th>Tasks</th>
<th>1* Free task</th>
<th>2* Manual task</th>
<th>3* $F_H$-$V_R$-$B_{L\perp}$ $K_{S\perp}$-$L_{i\perp}$</th>
<th>4* $F_V$-$V_R$-$B_{L\perp}$ $K_{S\perp}$-$L_{A\perp}$</th>
<th>5* $F_H$-$V_R$-$B_{L\perp}$ $K_{N\perp}$-$L_{A\perp}$</th>
<th>6* $F_V$-$V_S$-$B_{L\perp}$ $K_{S\perp}$-$L_{A\perp}$</th>
<th>7* $F_H$-$V_R$-$B_{H\perp}$ $K_{S\perp}$-$L_{A\perp}$</th>
<th>8* $F_V$-$V_K$-$B_{H\perp}$ $K_{S\perp}$-$L_{A\perp}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max. hor. force at the hands (N)</td>
<td>X</td>
<td>136</td>
<td>121</td>
<td>154</td>
<td>89</td>
<td>149</td>
<td>81</td>
<td>150</td>
<td>92</td>
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<tr>
<td>S.D.</td>
<td></td>
<td>23</td>
<td>30</td>
<td>23</td>
<td>28</td>
<td>26</td>
<td>35</td>
<td>27</td>
<td>21</td>
</tr>
<tr>
<td>Max. vert. force at the hands (N)</td>
<td>X</td>
<td>130</td>
<td>181</td>
<td>143</td>
<td>142</td>
<td>98</td>
<td>138</td>
<td>143</td>
<td>136</td>
</tr>
<tr>
<td>S.D.</td>
<td></td>
<td>26</td>
<td>39</td>
<td>25</td>
<td>30</td>
<td>89</td>
<td>26</td>
<td>51</td>
<td>27</td>
</tr>
<tr>
<td>Max. shear force at L5/S1 (N)</td>
<td>X</td>
<td>594</td>
<td>661</td>
<td>624</td>
<td>570</td>
<td>575</td>
<td>566</td>
<td>642</td>
<td>554</td>
</tr>
<tr>
<td>S.D.</td>
<td></td>
<td>63</td>
<td>86</td>
<td>62</td>
<td>77</td>
<td>65</td>
<td>75</td>
<td>83</td>
<td>69</td>
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<tr>
<td>Max. compression force at L5/S1 (N)</td>
<td>X</td>
<td>-2567</td>
<td>-3401</td>
<td>-3139</td>
<td>-3153</td>
<td>-3103</td>
<td>-3177</td>
<td>-3526</td>
<td>-2479</td>
</tr>
<tr>
<td>S.D.</td>
<td></td>
<td>1336</td>
<td>1327</td>
<td>1383</td>
<td>1205</td>
<td>1293</td>
<td>1374</td>
<td>1256</td>
<td>1241</td>
</tr>
<tr>
<td>$\int (F. \text{ comp. } \times \text{ dt}) (N \cdot s)$</td>
<td>X</td>
<td>2137</td>
<td>3503</td>
<td>2481</td>
<td>2043</td>
<td>2436</td>
<td>3776</td>
<td>2589</td>
<td>1671</td>
</tr>
<tr>
<td>S.D.</td>
<td></td>
<td>1207</td>
<td>1596</td>
<td>1247</td>
<td>1193</td>
<td>1428</td>
<td>2362</td>
<td>1058</td>
<td>1101</td>
</tr>
<tr>
<td>$\Delta$ time (max. comp. force-transition) (%$)</td>
<td>X</td>
<td>2.3</td>
<td>2.3</td>
<td>0.2</td>
<td>-0.7</td>
<td>0.8</td>
<td>3.4</td>
<td>1.6</td>
<td>5.8</td>
</tr>
<tr>
<td>S.D.</td>
<td></td>
<td>16.5</td>
<td>19.9</td>
<td>12.3</td>
<td>10.3</td>
<td>6.1</td>
<td>23.5</td>
<td>16.6</td>
<td>19.7</td>
</tr>
</tbody>
</table>

* $F_H$, $F_V$: hor., vert. force; $V_R$, $V_S$: rapid, slow vel.; $B_{L\perp}$, $B_{H\perp}$: low, high bed $K_S$, $K_N$: knee supported, non-supported; $L_{A\perp}$, $L_{T\perp}$: legs apart, together.

Sign. diff.: 3-4; 1-4; 1-6; 1-8; 2-3; 2-4; 2-6; 2-8.

Sign. diff.: 3-4; 3-5; 1-2; 2-3; 2-4; 2-5; 2-6; 2-7; 2-8.

Sign. diff.: 3-4; 3-5; 1-2; 1-7; 2-4; 2-5; 2-6; 2-8.

Sign. diff.: 4-8; 1-7; 2-8.

Sign. diff.: 4-6; 1-2; 1-6; 2-4; 2-8.

Not tested statistically.
Figure 5. Spinal loads at the lumbo-sacral joint as a function of movement time in an horizontal task (task no. 3 in figure 4).

3.1.3. Muscle activity: The EMG results are presented in table 3 for 12 subjects only, since several trials were not available for the remaining subjects due to problems encountered in data acquisition. The results are presented for the erectors, and for the left deltoid only, because of the high degree of similarity observed in the deltoids of the right and left arms. The data for the rectus abdominis and obliquus were obtained for three subjects only for purposes of validating the model.

The average EMG amplitudes were expressed as percentages of the maximal isometric contraction of the particular muscle. The predominance of deltoid activity generally occurred during the turning phase and may be due to the nature of the movement, the medial portion of the deltoid being more loaded during arm abduction as seen during the turning action. Maximal activity in the deltoids was observed to occur a little after the transition point, from 2·8% to 18·1%; it has already been mentioned that arm flexion seems to occur only after the hips start moving forwards.

For the three subjects and all the trials analysed, the rectus abdominis displayed low activity, average amplitude here for all tasks being 14·6% of maximal voluntary isometric contraction, the lowest average being 9·2% for the free task and the highest average, 19·2%, for the manual task. On the other hand, the obliquus muscle was considerably more active with an average amplitude over all tasks of 29·2% (range: 20·8% for the free task and 38·5% for the manual task). These muscles were less active still than the erectors muscles, which showed an average amplitude of 40·7%, with a minimum of 27·7% for the free task and a maximum of 53·3% for task 6.

3.2. Parameters of task execution

In this section, different execution parameters were analysed statistically and the following comparisons were made: direction of applied force, velocities, knee support, leg positions, heights of the bed. The free task and manual task were also compared with each other and with all other tasks.
Table 3. Electromyographic activities of the erectores and deltoid muscles for the different tasks (N=12).

<table>
<thead>
<tr>
<th>Tasks</th>
<th>Variables</th>
<th>1* Free task</th>
<th>2* Manual task</th>
<th>3* $F_{H-V_R-B_L}$ $K_S-L_A$</th>
<th>4* $F_{V-V_R-B_L}$ $K_S-L_A$</th>
<th>5* $F_{H-V_R-B_L}$ $K_N-L_A$</th>
<th>6* $F_{V-V_S-B_L}$ $K_S-L_A$</th>
<th>7* $F_{H-V_R-B_L}$ $K_S-L_T$</th>
<th>8* $F_{V-V_R-B_H}$ $K_S-L_A$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Total average amplitude, left deltoid (%)b</td>
<td>X</td>
<td>S.D.</td>
<td>49.8</td>
<td>33.9</td>
<td>45.1</td>
<td>74.8</td>
<td>38.6</td>
<td>76.8</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td></td>
<td></td>
<td>27.6</td>
<td>25.6</td>
<td>27.8</td>
<td>32.9</td>
<td>23.5</td>
<td>27.6</td>
</tr>
<tr>
<td></td>
<td>Total average amplitude, erectores (%)c</td>
<td>X</td>
<td>S.D.</td>
<td>28.0</td>
<td>45.5</td>
<td>37.8</td>
<td>46.9</td>
<td>40.3</td>
<td>42.7</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td></td>
<td></td>
<td>9.3</td>
<td>12.8</td>
<td>14.0</td>
<td>20.3</td>
<td>12.5</td>
<td>18.8</td>
</tr>
<tr>
<td></td>
<td>$\int$ EMG $\times$ dt., left deltoid ($\mu$V $\cdot$ s)$^d$</td>
<td>X</td>
<td>S.D.</td>
<td>196.1</td>
<td>124.6</td>
<td>132.4</td>
<td>182.6</td>
<td>114.0</td>
<td>305.6</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td></td>
<td></td>
<td>144.6</td>
<td>142.0</td>
<td>83.6</td>
<td>79.9</td>
<td>41.8</td>
<td>206.1</td>
</tr>
<tr>
<td></td>
<td>$\int$ EMG $\times$ dt., erectores ($\mu$V $\cdot$ s)$^e$</td>
<td>X</td>
<td>S.D.</td>
<td>80.2</td>
<td>134.4</td>
<td>85.4</td>
<td>79.5</td>
<td>97.0</td>
<td>119.4</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td></td>
<td></td>
<td>31.9</td>
<td>68.7</td>
<td>35.3</td>
<td>28.5</td>
<td>41.8</td>
<td>74.0</td>
</tr>
<tr>
<td></td>
<td>$\Delta$ time (ampl. max. erectores-transition) (%)$^f$</td>
<td>X</td>
<td>S.D.</td>
<td>18.1</td>
<td>-4.1</td>
<td>8.0</td>
<td>-0.5</td>
<td>11.1</td>
<td>-0.5</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td></td>
<td></td>
<td>24.0</td>
<td>14.4</td>
<td>15.3</td>
<td>10.4</td>
<td>12.0</td>
<td>15.6</td>
</tr>
<tr>
<td></td>
<td>$\Delta$ time (ampl. max. deltoid-transition) (%)$^g$</td>
<td>X</td>
<td>S.D.</td>
<td>11.1</td>
<td>18.1</td>
<td>8.7</td>
<td>49</td>
<td>16.3</td>
<td>2.8</td>
</tr>
<tr>
<td></td>
<td>S.D.</td>
<td></td>
<td></td>
<td>17.0</td>
<td>24.8</td>
<td>17.0</td>
<td>10.6</td>
<td>19.5</td>
<td>17.3</td>
</tr>
</tbody>
</table>

* $F_{H}$, $F_{V}$: hor., vert. force; $V_{R}$, $V_{S}$: rapid, slow vel.; $B_{L}$, $B_{H}$: low, high bed; $K_{S}$, $K_{N}$: knee supported, non-supported; $L_{A}$, $L_{T}$: legs apart, together.

$^a$ Not tested statistically.

$^b$ Sign. diff.: 3–4; 4–8; 1–4; 1–6; 1–8; 2–4; 2–6; 2–8.

$^c$ Sign. diff.: 3–4; 1–2; 1–4; 1–8.

$^d$ Sign. diff.: 3–4; 4–6; 4–8; 1–6; 2–6; 2–8.

$^e$ Sign. diff.: 4–6; 1–2; 2–3; 2–4; 2–7.
Muscular forces were much more pronounced and the horizontal task was associated with considerably less activity in the deltoids and slightly more activity in the erectors as observed from average EMG amplitude and the integral of EMG activities as a function of time.

From their responses to the questionnaire about which mode of execution appeared easier, the nursing aides favoured the horizontal task. However, their perceptions may have been influenced by their free task which was also a horizontal task in all cases.

3.2.2. Velocities: Two tasks were compared which differed only in the speed of execution. This factor could not be rigorously controlled however and it was necessary to examine velocities a posteriori for all subjects. They were instructed to use a slow and a rapid speed consistent with typical situations encountered in the hospital. The velocity was measured at the hands and reflected the velocity applied to the patient. The following factors were held constant: vertical action, knee supported on the bedside, legs apart, low bed (tasks 6 and 4: slow and rapid).

The tasks differed kinematically and the slow task lasted longer and showed a lower average velocity (about 30%) and a lower average acceleration (about 50%). These results were expected and they indicate that the tasks differed along the main factor examined. Since the motion lasted longer, the time-based effects of spinal loading were considerably higher, about 85%, as were the time effects of muscle activity, which increased 65% in the deltoids and 50% in the erectors.

When questioned about their perceptions of the relative difficulty of the task, 11 out of the 15 subjects favoured the rapid task but did mention that the type of patient (body weight, state of health) is the most important factor. Generally, they will tend to work faster when the patient is heavy and if his/her illness does not prohibit more rapid movement.

3.2.3. Knee support on the bedside: Two tasks were compared where the knee could be supported or non-supported on the bedside (tasks 3 and 5). These tasks were executed horizontally, at rapid velocity, with legs apart and the bed in the low position. Knee support was provided by the force platform.

These tasks differed substantially in terms of the kinematic factors: when the knee was not supported, the movement lasted longer and the velocity was slightly lower but the movement had to be initiated with greater acceleration. The back was also more inclined at the start and motion was executed with more segmental amplitude, reflected in the backward motion of the hips. There was little to discriminate between the tasks in terms of the external and internal forces: when the knee was unsupported, the nursing aides experienced smaller vertical forces at the hands; maximum shear forces were also smaller but these differences were not important (about 7%); finally EMG activities were similar in both tasks.

All the nursing aides preferred to work with the knee supported against the bedside, a position they associated with better body balance. Even if our results do not present scientific evidence in favour of either task, use of knee support should probably be recommended. In the course of the experiment, subjects were often observed to be off-balance in the task where the knee support was absent and in fact several trials had to be repeated.
3.2.4. *Leg position:* The effects related to the symmetry or asymmetry of the legs in the antero-posterior direction were examined (tasks 7 and 3). These two tasks were executed in the horizontal direction, at rapid velocity, with the knee(s) supported and the bed in low position. In the asymmetric position, the subject was free to choose how far apart to set the front and rear foot and the average distance was 22.0 ± 8.3 cm.

There were no statistical differences found in the kinematics and the kinetics of the tasks. Opinion was divided among the nursing aides: six subjects preferred the asymmetric position, four preferred the symmetric and the others were indifferent. This factor does not therefore seem to be very important.

3.2.5. *Height of the bed:* The two tasks considered differed in the height of the bed: there was a difference of 10 cm, the higher position being at about hip level. The two tasks were controlled for the vertical component of the applied force, rapid execution velocity, leg asymmetry and knee support (tasks 4 and 8: low and high bed).

Of all the experimental factors examined, height of the bed was the most important, as reflected in the extent of the differences in task execution. When the bed was in the higher position, the trunk was nearer the vertical (30° vs. 38°) and consequently maximal muscle moment at L5/S1 was smaller, which produced smaller maximal compression forces (21%). The EMG measurements also revealed that the arms were more active, displaying greater average EMG activity in the deltoids (about 20%) and greater time effects of these activities (about 40%).

An analysis of nursing aide perceptions also revealed that the majority preferred to work with the bed in a higher position. This position may reduce the loads on the spine but with it the shoulder joints might be subjected to more stress, as is suggested by the higher muscle activity.

3.2.6. *The free task with the piqué:* This task represented the normal way the nursing aide handled her patients. Kinematically, it was the task associated with the longest duration and the slowest initial acceleration. In this task the trunk was maintained nearer the vertical. The characteristics of this task were similar to the horizontal task in terms of patterns of execution and kinetic factors. As a matter of fact, when we examined the external and internal forces at L5/S1, this free task was found to be one of the less strenuous. Also, the electromyographic data revealed that this task involved less strain on the deltoids and/or the erectors than did vertical tasks 4, 6, 8. Among horizontal tasks 3, 5, 7, this task appears optimal; it is apparent that through years of practice, the nursing aides may well have developed a more efficient way of working. However, this was not systematic throughout all subjects and mention must be made that the slower execution may have influenced the result but still it is interesting to notice these adjustments selected by the nursing aides.

3.2.7. *The manual task without the piqué:* This task is not a customary one; this movement may be encountered in practice but rarely in the manner examined in this study. The subject will not usually translate a patient in one single action, but will usually break down the movement by translating first the lower part of the body and subsequently the upper part of the body. This task was therefore only used as a basis for comparison.

Analysis of the kinematic factors revealed that this task was accomplished with the lowest velocity. The most striking factor was the trunk flexion: the trunk was near horizontal on initiation of the motion, with the differences in inclination from the other
tasks varying between 27° and 40°. With this task very large vertical forces were exerted by the nursing aide on the patient. The combination of large vertical forces and high trunk flexion may have caused the large loads experienced on the spine. The analysis of EMG activities revealed that this task was less demanding on the deltoids but the erectors muscles were very highly loaded.

4. Discussion

The most important findings in this study relate to the location of maximal compression forces and maximal EMG activity. In more than 80% of the cases, the maximal compression forces occurred within ±10% of the transition point where the hips initiated a change of direction in the motion. The same observation applies to maximal EMG activities. It has already been mentioned that when the hips initiated their change of direction, the arms were often seen to start moving, continuing translation of the patient to the side of the bed. This arm action was observed from the kinematics, where the arms were seen to go into flexion and hyperflexion at the beginning of the turning phase; it is only later during the turning phase that the arms were seen to extend. A conclusion, however, based solely on muscle activity in the medial portion of the deltoid may present some drawbacks, since this muscle is particularly active during arm abduction, a type of movement occurring when the patient is turned.

Many safety programmes for nursing aides often stress use of the lower limbs and a counter-weight approach, emphasizing the force of gravity, the purpose of these suggestions being to minimize action in the arms and back. For all the nursing aides in this study, whose work experience varied between 1 and 15 years, arm action was far from absent. This action, based on EMG data on the deltoids and hand forces, was not substantial in the first phase, however, and it is possible that the subjects may have favoured a counter-weighted approach in this phase. There were, very large forces at the end of the first phase or at the beginning of the second, defined as the turning phase. These large forces were generally associated with higher compression forces at L5/S1.

These findings may also suggest that the concept of risks for the spine might be associated not only with the weight of the body handled during the work activities but also with a pattern of motion involving changes of direction. Changes of direction in the trunk motion when handling large body weights such as are encountered in the tasks of pulling and turning patients, that is, changes from a backward to a forward direction, may be hazardous for the spine. It might be more appropriate to break the movement down into several units linked by slight pauses rather than to use a continuous motion with changes of direction. Further experimental verification of this hypothesis must be done before any general inferences can be warranted.

The intensity of the loads on the spine observed in the tasks under consideration was not excessively high, recognizing that intra-abdominal force was omitted in the model. The loads varied on the average from 2479 N to 3526 N, in maximal compression. In line with the NIOSH proposal that all loads below 3430 N should not require assistance, these tasks do not appear to present any particular problem but it must be borne in mind that several individuals will sometimes substantially exceed these values, as was reflected by the high standard deviations in our study. A previous study on lifting patients from a chair led to considerably higher loads, varying on the average from 5744 to 7951 N, depending on the method used (Gagnon et al. 1986). Ekholm et al. (1982), with a similar but quasi-static model, showed compression forces to vary between 3461 and 4425 N for different techniques of lifting a 12.8 kg load.
Shear loads have received little experimental attention in the literature: a comparison with the previous Gagnon et al. (1986) study showed that the ratios of maximal shear forces to maximal compression forces were about the same for lifting patients as for translating-turning movements, that is, were between 12 and 17% for the different methods of lifting and between 18 and 23% for the different turning tasks. It was expected that shear forces would increase more substantially in this type of movement, involving pulling and turning the patient in bed, given the main direction of the effort. That was not the case and it is possible that the subject tends to assume a trunk position and to orient his effort so as to minimize shear forces.

Use of a dynamic approach may have been justified when compared to a quasi-dynamic approach (external forces recorded but inertial forces set at zero). Comparisons for two subjects performing the 8 tasks revealed that maximal compression forces increased by an average of 8% and maximal shear forces by an average of 2%. With one exception, there was a systematic increase in compression forces with the dynamic approach, whereas no consistent patterns showed up for shear forces. These results were different from those reported by McGill and Norman (1985) for a lifting task: they found that resulting L4/L5 moments were 25% smaller when a dynamic approach was used. These differences may be attributed to the different nature of the movements analysed in both studies.

Our model was developed to predict spinal loads during extension efforts and not during flexion efforts, since the rectus abdominis was not included in this model. It was therefore necessary to verify if this type of flexion efforts occurred during the experiment; it turned out that net flexor muscle moments at L5/S1 were very rarely experienced during the experiment and those trials were therefore eliminated. The behaviour of the efforts in extension was corroborated by the very low level of activity encountered in the rectus abdominis among the subjects examined. Usually the subject assumed a position which favoured extension by lowering his body during the pulling and by raising his body when turning or pushing the patient.

In this study, the hypothesis of symmetry of arm actions has been assumed. In general, the EMG patterns of the deltoids were similar for both arms but there were dissimilarities in some subjects, an indicator that asymmetry may have been present in them. However, this was rather subject specific than task specific. The development of tridimensional models of spinal loadings will greatly help the analysis of movements where asymmetry is suspected.

Analysis of the different execution parameters has demonstrated that the horizontal pull on the patient increased maximal shear forces and muscular activity in the erectors but reduced activity in the deltoids. The question remains as to which strategy should be promoted: reduction of spinal loads or reduction of loads on the shoulder joint. This question also arose when we analysed the effects of bed height in the vertical tasks: when the bed was high there was a substantial reduction in compression loading on the spine, but the arm abductors, in other words, the deltoids, were considerably more active. We should never lose sight of the possibility of transferring the problem to other joints when we make recommendations based on analyses that have been performed too narrowly. We should probably be aiming at optimizing joint loading.

The effects of velocities were more easily understood: a rapid velocity should be encouraged as it is associated with considerable reduction in spinal loading and EMG activity analysed for their time effects. On the other hand, knee support and leg position did not appear to affect the internal forces to any significant degree.
This study was restricted to analysis of internal loads but an investigation of mechanical work and energy changes may be imperative when examining work operations. Recommendations with regard to spinal loads must be viewed in relation to the energy cost required for the task if long-term performance is desired. The data on mechanical work have been reported in another study (Gagnon et al. accepted for publication). The most important finding concerns the lack of agreement between judgments of optimality based on work-energy and on minimization of loads on specific joints. Which system does the worker choose? Which is best from the safety point of view? The question has yet to be answered.

Conclusion

In order to reduce spinal loads it is recommended that the action of pulling and turning the patient over in bed should be accomplished with the forces being exerted in the vertical direction, with the bed in the high position (hip level) and with rapid motion if the patient's condition permits.

One should be aware of the possibility of transferring the problem from the spine to other joints. The change of direction in the trunk motion may in itself present risks when heavy weights are handled.

Dynamic segment analyses improved prediction by about 10%, predicting generally heavier loads than are estimated from a quasi-dynamic model.

Acknowledgments

This research was supported with a grant from the Institut de recherche en santé et en sécurité au travail du Québec (N/D RS-84-11).

La manutention de patients dans le lit avec un piqué (un tissu matelassé à l'épreuve de l'eau placé sous le patient) est associée à une incidence élevée de risque pour la colonne où, en particulier, l'activité consistant à tirer et retourner le patient avec le piqué représente le plus haut risque. Quinze préposées aux malades furent évaluées pour les forces de compression et de cisaillement à l'articulation L5/S1 et pour des activités musculaires sélectionnées au niveau du tronc et des épaules. Les données furent obtenues à partir de films, de plate-formes de force et d'enregistrements électromyographiques; des analyses segmentaires dynamiques furent effectuées pour calculer les forces de réaction en L5/S1 et un modèle planaire à un seul muscle équivalent fut utilisé pour estimer les charges internes. Différents paramètres d'exécution furent examinés incluant la vitesse d'exécution, la hauteur du lit, la direction des efforts, la position des jambes et le support au genou. Une tâche 'libre', et une tâche manuelle n'impliquant pas l'usage du piqué, furent aussi étudiées. Des recommandations sont effectuées pour réduire le chargement sur la colonne. Les résultats suggèrent aussi qu'un changement de direction dans le mouvement du tronc peut présenter des risques lorsqu'il est associé à la manutention de charges lourdes. De plus un transfert de problème d'une articulation particulière à d'autres articulations peut se présenter.

Patienten unter Benutzung eines 'Pique' (ein wasserdiches gepolstertes Tuch, das unter den Patienten gelegt wird) umzulagern, ist mit einem erhöhten Risiko für die Wirbelsäule verbunden, insbesondere Ziehen und Drehen des Patienten mit Hilfe des 'Pique' stellen das höchste Risiko dar. Fünfzehn weibliche Schwesternhelferinnen wurden bezüglich der Druck- und Scherkräfte am L5/S1 Gelenk und ausgewählter Muskelaktivitäten im Rücken- und Schulterbereich untersucht. Filme, Messungen mit Kraftmeßplatten und EMG-Aufzeichnungen lieferten die Daten, biomechanische Analysen der einzelnen Körperteile wurden ausgeführt, um die Reaktionskräfte bei L5/S1 zu berechnen und ein Modell der Wirbelsäule wurde benutzt, um die inneren Kräfte abzuschätzen.

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