AN ABSTRACT OF THE DISSERTATION OF

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The dependent transfer of a traveler with disabilities on board an aircraft places the transferors, as well as the traveler, at risk of disabling lower back injury. This risk is increased on board an aircraft due to the confined space in which the transfer takes place. The purpose of the current study was to determine the influence of the spatial constraints commonly found on aircraft, transferee size, and transfer direction on kinetic and kinematic risk factors associated with increased risk of low back disorders, during dependent transfers on an aircraft. Also of interest was the perceived task difficulty of the transferors, and what influence this may have on the safety of the traveler being transferred.

Thirty-three pairs of apparently healthy men and women worked together to perform two-person dependent transfers between a wheelchair and an airplane seat in a laboratory simulation of an aircraft interior using two sizes of anthropometric dummies. A three-way repeated measures ANOVA was used to determine the influence of spatial constraints, transferee size, and transfer direction on known kinematic and kinetic risk factors for low back injury during lifting tasks.

Results indicate that the constraints significantly influenced lumbar motion of the front transferor, due to the need to reach around the seat. As a result, load moment arm was also significantly increased. Lumbar motion of the rear transferor was most influence by the size of the transferee. Flexion angles for both transfer positions were found to be greater than those known to result in high risk of low back disorder. Transferee size significantly increased lumbar loading of the rear transferor. Shorter transferors were found to experience increased lumbar motion during transfers performed from the rear position. Although constraint and transferee size influenced the perception of task difficulty, the acceleration experienced by the transferee remained unchanged. The variability associated with this measure indicates transferor safety is influenced by the persons performing the transfer.

These results indicate that transferee size, the spatial constraints imposed by the aircraft interior, and transferor height all affect the risk of low-back injury during dependent transfers on an aircraft.

BIOMECHANICS OF DEPENDENT TRANSFERS ON AN AIRCRAFT

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I understand that my dissertation will become part of the permanent collection of Oregon State University libraries. My signature below authorizes release of my dissertation to any reader upon request.

Brian K. Higginson, Author

CONTRIBUTIONS OF AUTHORS

Dr. Pavol contributed to the study design and assisted with data collection, development of data analysis programs, statistical analysis and interpretation. Chris Schafer designed and fabricated the force plate mounted in the aircraft seatback that allowed for the collection of force data applied to the seatback during the transfers. Lisa Welsh contributed to the development of kinematic data analysis programs, as well as made significant contributions during the data collection process.

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1. INTRODUCTION

In recent years, air travel has become an increasingly popular and necessary means of travel for both business and pleasure. An estimated 768 million passengers will be flying this year alone, with more than 1 billion predicted by the year 2015 (FAA, 2007). With the population of people with disabilities expected to increase by 30.9 million between 1997 and 2030 (US Census Bureau, 2001), the need to accommodate this population during air travel is apparent. Presently, travelers with disabilities spend $3.3 million per year on travel, a number that could be increased to as much as $27 billion per year if the travelers are properly accommodated (FAA, 2007). Unfortunately, the vast majority of disabled travelers (84%) encounter obstacles when dealing with airlines (Open Doors Organization, 2005), discouraging many from traveling unless necessary.

An issue of major concern to travelers with disabilities during air travel is the transfer to and from mobility aids during the boarding and deplaning processes. The narrow aisles found on board an aircraft do not accommodate personal mobility aids, requiring travelers to be transferred from their personal mobility aid to an aisle chair prior to boarding. Once on board the aircraft, they must once again be transferred from the aisle chair to their assigned seat. This results in a minimum of four transfers per flight, with the potential for many more if use of the lavatory is required or if the trip includes connecting flights. Each of these transfers puts the traveler, as well as the persons performing the transfer, at risk of injury. During these transfers, the traveler is reliant on the ability of the transferors to safely execute the task, and could potentially suffer shoulder, back, or other injuries if dropped or mishandled. The transferors are also at a high risk of low back or other musculoskeletal injury while performing these transfers. Obviously, there is a need to understand and reduce these risks.

In order to understand the mechanisms responsible for low back pain or injury, we must be able to accurately assess the loading that occurs at the most common site of injury, the lower lumbar spine, during whole-body, free, dynamic lifting conditions. An accurate evaluation of spinal loading is necessary in order to compare loads imposed upon the lumbar spine to known tolerance limits. The evaluation of loads on the lumbar spine is a rather difficult task. Unfortunately, at the present time, there is no feasible method whereby such data can be measured directly. Direct measurement techniques, such as intra-discal force transducers, are expensive and dangerous due to their invasive nature, particularly when motion is involved.

For this reason, several biomechanical analysis techniques (models) have been developed that are capable of predicting loads on the lumbar spine based on Newtonian laws of physics. These techniques have been developed to objectively quantify factors known to precipitate low back pain or injury, including kinematic, kinetic, and workplace variables. Kinematic variables associated with increased risk of injury include lumbar flexion, lateral bending, and twisting angles, and lumbar flexion, lateral bending, and twisting velocities. (Marras et al., 1993; Norman et al., 1998; Punnett et al., 1991; Fathallah, 1998). Kinetic variables include forces crossing the lower lumbar spine (compressive, anterior-posterior shear, and medial-lateral shear forces) and lumbar moments about the three principle axes of motion (flexion, lateral bending, and twisting moments) (Herrin et al., 1986; Marras et al., 1993). Although an increase in any single kinematic or kinetic variable leads to an increase in low back injury risk, a concomitant increase in multiple factors has been found to be most predictive of injury risk (Granata & Marras, 1999; Marras et al., 1993). In addition to these biomechanical variables, greater external loads in the hands (Norman et al., 1998; Waters et al., 1993; Snook & Ciriello, 1991) and longer load moment arms (horizontal distance of the load from the spinal segment during a lift) have been found to be perhaps the most sensitive predictors of increased risk of injury (Marras et al., 1993; Anderson et al., 1976).

Currently there exists a paucity of information regarding the prevalence of low back injuries during dependent transfers of travelers with disabilities on board aircraft, and no quantitative research exists addressing the potential for injury while performing such a task. Due to the high prevalence of occupational back injuries associated with patient-handling tasks in clinical and assisted care settings, research in this area is more prolific. Most patient handling tasks can be broken down into three categories: lifting/transferring, repositioning, and turning (Skotte et al., 2002). Therefore, the current literature on transfers typically focuses on comparison of injury risk between the different patient-handling tasks (Schibye et al., 2003; Daynard et al., 2001; Skotte et al., 2001, 2002), the influence of different techniques on the same task (Varcin-Coad & Barrett, 1998; Winkelmolen et al., 1994), comparison of one- and two-person transfers given the same task (Marras et al., 1999a; Garg et al., 1991), or the difference between manual transfers and the use of mechanically-assisted transfer devices for the same task (Ulin et al., 1997; Garg et al., 1991).

Despite the wide scope of the literature, information regarding potential injury risk during two-person dependent transfers is limited to very few patient-handling studies (Daynard et al., 2001; Marras et al., 1999a; Ulin et al., 1997; Garg & Owen, 1993; Garg et al., 1992;Garg et al., 1991; Owen & Garg, 1991) and even fewer two-person manual materials handling studies (Karwowski, 1998; Karwowski & Mital, 1986; Karwowski & Pongpatanasuegsa, 1988; Marras et al., 1999a). Of the relevant literature, there seems to be a general consensus that a patient transfer task has the potential to result in a high risk of low back injury. The estimated loads acting on the spine of the transferor have been found to approach, or sometimes exceed, known tolerance limits.

However, although the patient handling literature provides insight into the spinal loadings during a dependent transfer, the transfers on board an aircraft differ from those studied to date in several, potentially important respects. First, unlike the transfers performed in most other settings, dependent transfers on board aircraft are typically performed using a two-person “front-and-rear” technique. In this technique, the front and rear transferors grasp the transferee beneath the knees and through the arms, respectively, and lift together to shift the individual laterally (Pelosi & Gleeson, 1988). With the possible exception of Ulin et al. (1997), who looked at transfers to and from a bed, the front and rear transfer technique used on board an aircraft has not previously been investigated.

Second, the effects of surrounding spatial constraints while transferring a person with disabilities have yet to be studied. Previous research has focused primarily on the risk of injury during patient transfer tasks performed in clinical or assisted care settings. However, these settings differ considerably from the conditions on board an aircraft. The interior of an aircraft imposes rigid spatial constraints that are probably the most extreme found in any patient handling environment. The constraints imposed by the interior of an aircraft, including surrounding rows and aisles of seats, overhead bins, and high seat backs, could potentially result in an increased risk of low back injury by altering known kinematic and kinetic risk factors. Should this be the case, future redesign considerations may be warranted. It is possible that increases in aisle or row spacing may decrease risk of injury to more acceptable levels.

Other factors that have the potential to influence the risk of low back injury during dependent transfers on board an aircraft include the size of the individual being transferred, as well as the direction of the transfer (inboard vs. outboard). Studies have indicated that heavier loads typically result in greater compressive forces and load moments on the lumbar spine. However, as stated earlier, none have examined the transfer technique on board an aircraft, particularly where the rear transferor must work over or around a high seat back. Because of the seating geometry, especially the difference in the height and depth of the backs of the aircraft seat and aisle chair, the potential also exists for the risk of low back injury to change as a function of transfer direction.

Finally, most of the current literature investigating patient handling tasks tends to focus on the risk of injury to the transferors. However, during dependent transfers, the transferee is also at risk of injury should he or she be dropped or mishandled. Any factor preventing a safe and controlled transfer by the transferors could potentially increase the risk of serious injury to the person being transferred. There is little information quantifying this injury risk to the transferee.

The purpose of this study is therefore to determine how the constraints imposed by the interior of an aircraft, the transferee size, and the direction of transfer affect risk factors for injury during dependent transfers of travelers with disabilities on board an aircraft. The study begins by investigating the influence of the constraints imposed by the interior of an aircraft, the transferee size, and the direction of transfer on known risk factors for low back injury to transferors during dependent transfers on board an aircraft. Chapter 2 provides a kinematic analysis of both the front and rear transferors performing a dependent transfer in a simulated aircraft environment. Chapter 3 utilizes a kinetic analysis focused on the rear transferor to determine the influence of the aforementioned factors on lumbar compression forces and moments during the transfers. Chapter 4 then considers the potential for injury to the individual being transferred by quantifying the peak vertical acceleration experienced by the traveler during the transfers. In this chapter, the effects of spatial constraint, direction of transfer, and transferee size on perceived task difficulty of the individuals performing the transfer are also investigated, as well as whether these perceptions of task difficulty influence the risk of injury to the traveler.

Given this information, it is hopeful that procedural or environmental factors can be modified to reduce the risk of injury to travelers with disabilities and those who assist them during dependent transfers on board an aircraft.

KINEMATICS OF DEPENDENT TRANSFERS ON AN AIRCRAFT

Brian K. Higginson, Lisa R. Welsh and Michael J. Pavol

2.1 INTRODUCTION

The ability for people with disabilities to travel by commercial aircraft is important for their full participation in society, but poses challenges to airline and airport personnel. An area of great concern, from an ergonomics standpoint, is the process of transferring a traveler with disabilities between a wheelchair and an aircraft seat when the traveler is unable to do so. These dependent transfers on board aircraft are typically performed using a two-person "front-and-rear" technique, in which the front and rear transferors grasp the transferee beneath the knees and through the arms, respectively, and lift together to shift the individual laterally (Pelosi & Gleeson, 1988). Such transfers place the transferors at risk of a disabling injury to the lower back, a risk that needs to be reduced.

Although statistics on the incidence of low-back injury during transfers on board aircraft are not readily available, in clinical and assisted care settings, lifting and transferring patients have been identified as frequent precipitating factors or causes of low-back disorders among nurses (Ulin et al., 1997; Winkelmolen et al., 1994). In fact, nurse's aides have the highest incidence of disabling back injuries in the USA (Jensen, 1987). Winkelmolen et al. (1994) suggest several reasons why lifting a patient may be much more strenuous than lifting in industry: the shape and weight of a patient do not conform to industrial recommendations, the spatial conditions interfere with the working posture, and most nurses are women. Despite the high rate of low-back injuries among patient handlers, there has been limited biomechanical study aimed at quantifying and reducing the risk of injury during a transfer task (Garg et al.,1991a; Marras et al., 1999a).

The dependent transfer of a traveler with disabilities on board an aircraft poses unique challenges to the transferors that may further increase the risk of injury. The interior of an aircraft imposes rigid spatial constraints that are probably the most extreme found in any patient handling environment. Constraints include the adjacent rows and aisles of seats, the overhead bins, and the high back of the aircraft seat itself. To date, the effects of these constraints on risk factors for low-back injury to transferors have not been determined. Nor is it known whether the effects of these constraints differ as a function of transfer direction (i.e. wheelchair-to-seat or seat-to-wheelchair). Also of consideration, transfers performed with heavier patients in clinical and assisted care settings have been shown to result in an increased risk of low-back injury. The relative influence of the size of the individual being transferred on risk factors for low-back injury during transfers on board an aircraft is not known, however. By understanding these different effects, appropriate means of reducing the risk of injury may be developed.

Established associations exist between the kinematics of lifting tasks and the risk of occupationally-related low-back disorder (Marras et al., 1995). The purpose of this study was therefore to determine the effects of the spatial constraints found on board an aircraft, the direction of transfer, and the size of the individual being transferred on kinematic risk factors for low-back disorders during the dependent transfer of travelers with disabilities between a wheelchair and an aircraft seat. Risk factors considered include greater peak trunk angles, faster trunk motions, and larger load moment arms (Marras et al., 1995).

2.2 METHODS

Study participants consisted of 33 pairs of men (*n* = 42) and women (*n* = 24) between the ages of 18 and 40 years, each pair consisting of a front transferor and a rear transferor. The mean ± SD age, mass, and height were 24.1 ± 6.4 years, 68.8 ± 13.3 kg, and 172.0 ± 9.4 cm, respectively, for the front transferors, and 23.6 ± 4.2 years, 78.0 ± 13.7 kg, and 178.0 ± 7.7 cm for the rear transferors. Health history questionnaires were administered to screen participants for contraindications to the level of physical exertion required to perform a dependent transfer. Eligible subjects consisted of healthy individuals who were free from pre-existing back disorders or injuries and who had regularly been resistance training or the equivalent over the preceding month. Subjects were not required to have previous experience performing dependent transfers. Institutional Review Board approval was obtained and subjects provided written informed consent prior to participation.

Subjects worked in pairs to transfer an anthropometric dummy between an aircraft aisle wheelchair and an airplane seat. The two-person dependent transfers were performed in a laboratory simulation of an aircraft interior using the “through arm” transfer technique (Pelosi & Gleeson, 1988). In this technique, the rear transferor reaches beneath the shoulders of the transferee from behind, grasps the forearms, and secures them against the chest. The front transferor reaches under the thighs, grasping the transferee just above the knees. The transferors then synchronously lift and shift the transferee sideways upon a verbal three-count initiated by the rear transferor. Transfer position (front or rear) was self-selected by the subjects and was maintained for the duration of testing.

A standard economy-class airplane seat (Boeing, Seattle, WA) was mounted to the floor with the armrest raised. An aircraft aisle wheelchair (AisleMaster 8000, Columbia Medical, Pacific Palisades, CA) was placed adjacent to the aircraft seat and secured in place (Figure 2.1). A small anthropometric dummy (Simulaids, Saugerties, NY), representative of a 50th percentile woman (mass: 57 kg; height: 165 cm), and large (First Technology Safety Systems, Plymouth, MI) anthropometric dummy, representative of a 50th percentile man (mass: 78 kg; height: 178 cm), served as the transferee during all transfers.

Removable frames, constructed from electrical conduit, were used to simulate the spatial constraints imposed by the aircraft interior (Figure 2.1). These constraints included the surrounding rows and aisles of seats and the overhead bins and they were placed at locations approximating those found on a typical single-aisle aircraft. Aisle width was set at 43.2 cm, row pitch (distance from the front of one seat to the front of the next seat) was 78.4 cm, and overhead bins were located at a height of 162.5 cm above the floor and 30.5 cm outboard from the inboard (aisle) edge of the aircraft seat.

Subjects were instructed on proper transfer technique and, after a warm-up, were allowed two practice transfers prior to testing in order to familiarize themselves with the transfer task. The subjects then transferred each dummy once for each transfer direction (outboard: wheelchair-to-seat, and inboard: seat-to-wheelchair) under constrained and unconstrained conditions, for a total of eight trials. Trials were blocked by constraint, and within constraint, by dummy size. Blocks, and trials within blocks, were counterbalanced to account for potential order effects. At least two minutes of rest was provided between trials.

Three-dimensional body segment orientations were recorded during each transfer by using an eight-camera motion capture system (Vicon, Lake Forest, CA) to track 74 reflective markers attached to the subjects and dummy. Marker trajectories were sampled at 60 Hz and low-pass filtered using a 4^{th}-order zero-lag Butterworth filter with a cut-off frequency of 8 Hz, as determined by residual analysis of the raw marker position data (Winter, 1990). The recorded marker locations, along with anthropometric measurements and the relative marker positions during a trial of quiet standing, were used to determine the three-dimensional orientations of the pelvis and trunk during each trial. The position of a lumbar joint center located within the spine at the L_3/L_4 level was similarly determined. The flexion-extension, lateral bending, and twisting angles of the trunk relative to the pelvis were computed based on the corresponding Cardan rotation sequence. Angular velocities were then computed by numerical differentiation of the Cardan angles. Finally, the load moment arm was computed as the horizontal distance between the lumbar joint center and either a marker on the chest of the dummy for the rear transferor, or the point directly between both wrists for the front transferor.

Dependent variables extracted for each transfer included the peak lumbar angles (flexion, bending, and twisting), peak lumbar angular velocities (flexion, extension, bending, and twisting), and maximum load moment arms of the front and rear transferors. Left and right bending or twisting were considered to be equivalent (i.e. the absolute value was used) in the determination of the corresponding peak angles and angular velocities.

Three-way repeated measures analyses of variance were used to test the effects of spatial constraints, transfer direction, and transferee size on each dependent variable, with the front and rear transferor analyzed separately. Paired t-tests with a Bonferroni correction were used to investigate significant interaction effects. Systematic influences of the body height of the transferor on each dependent variable were assessed through linear regression. The mean value across transfers was used for each subject in the regression analyses. Effects were considered significant at $\alpha = .05$. Calculation of descriptive statistics, tests of model assumptions, and statistical analyses were performed using SPSS version 13.0 (SPSS, Chicago, IL).

2.3 RESULTS

2.3.1 Front Transferor

The spatial constraints imposed by the adjacent rows and aisles of seats had a significant effect on the peak kinematics of the front transferor (Table 2.1). The presence of the spatial constraints resulted in greater peak lumbar flexion ($p < 0.001$), lateral bending, and twisting ($p = 0.001$) angles of the front transferor. Post-hoc analysis of a constraint by direction interaction ($p = 0.049$) indicated the effect on lateral bending angle was dependent on the direction of transfer; peak lateral bending angle was 2.5 degrees ($p = 0.001$) and 4.3 degrees ($p < 0.001$) greater in the presence of constraints for inboard and outboard transfers, respectively (Table 2.2). The presence of the spatial constraints also resulted in a peak lumbar extension velocity that was 5.5 deg/s slower during the transfer ($p = 0.001$). Furthermore, the peak load moment arm of the front transferor was greater in the presence of the spatial constraints. The size of this effect depended on the size of the transferee (size x constraint interaction; $p = 0.006$); the load moment arm was 12.9 cm ($p < 0.001$) and 9.9 cm ($p < 0.001$) greater under constrained conditions for the smaller and larger transferee, respectively (Table 2.3).

Effects of transfer direction were limited to the peak lumbar flexion angle and flexion velocity (Table 2.4). The effect of transfer direction on lumbar flexion angle depended on transferee size (size x direction interaction; $p = 0.008$). Peak lumbar flexion angle was greater for inboard than for outboard transfers of the smaller dummy ($p < 0.001$), whereas direction had no effect for the larger dummy (Table 2.5). Similarly, the effect of transfer direction on lumbar flexion velocity depended on the constraint conditions (constraint x direction interaction; $p = 0.026$). In the presence of the spatial constraints, peak lumbar flexion velocity was greater during outboard transfers ($p = 0.010$), whereas transfer direction had no effect when the constraints were absent (Table 2.6).

Transfers of the larger transferee were associated with greater peak lateral bending angles ($p = 0.036$) and greater peak lumbar flexion velocities ($p = 0.010$; Table 2.7). In contrast, the peak load moment arm of the front transferor was 2.8 cm smaller for the larger transferee than for the smaller transferee during transfers in the presence of the spatial constraints (Table 2.3). Peak load moment arm was unaffected by transferee size in the unconstrained condition. Finally, there was a three-way interaction between the effects of transferee size, constraint, and transfer direction on the peak lateral bending velocity ($p = 0.045$); however, none of the simple effects of these factors were significant in the post-hoc analysis.

A significant linear relationship was found between the body height of the front transferor and the load moment arm (Figure 2.2), with taller transferors tending to have larger load moment arms ($r^2 = 0.29$; $p = 0.001$). Body height did not influence any of the other dependent measures in the front transferor.

2.3.2 Rear Transferor

The presence of the spatial constraints imposed by the adjacent rows and aisles of seats during dependent transfers resulted in few changes in peak lumbar kinematics of the rear transferor (Table 2.8). Peak lumbar flexion was minimally greater by 1.5 degrees in the constrained condition ($p = 0.044$). Although the direction of transfer also only influenced one dependent measure, the magnitude of the difference between conditions was proportionately much greater than that observed with constraint (Table 2.9). The peak lumbar flexion velocity of the rear transferor was, on average, 14 deg/s faster during outboard transfers ($p < 0.001$). Dependent transfers performed with the large transferee resulted in larger peak lateral bending angles ($p = 0.003$) and velocities ($p = 0.033$), larger peak twisting velocities ($p < 0.001$), and a larger load moment arm ($p < 0.001$; Table 2.10). In contrast, transfers of the smaller transferee were associated with greater peak flexion angles ($p < 0.001$; Table 2.10) in the rear transferor.

Body height had a substantial influence on the dependent measures in the rear transferor. Taller transferors tended to have lower peak lumbar twist angles (Figure 2.3), slower lateral bending velocities (Figure 2.4), and slower twisting velocities (Figure 2.5). As with the front transferor, greater peak load moment arms were associated with taller transferors (Figure 2.6). Also of note, the load moment arm of the rear transferor was, on average, nearly 17 cm greater than that of the front transferor.

2.4 DISCUSSION

These results indicate that the spatial constraints imposed by the interior of an aircraft significantly influence lumbar kinematics and the load moment arm during a dependent transfer of a traveler with disabilities. For the rear transferor, the presence of the spatial constraints resulted in a greater peak lumbar flexion angle. The constraints imposed by the adjacent seats appeared to have a more pronounced effect on the front transferor, negatively affecting lumbar flexion, lateral bending, and twisting, as well as the load moment arm.

Although the novelty of the transfer task investigated in the current study precludes direct comparison of the results to those of others, patient transfer tasks in which both transferors lift from the front of the transferee have shown similar lumbar kinematics. Transfers of patients from a wheelchair to a toilet using this technique have resulted in flexion angles of 53-68 degrees, lateral bending angles of 11-13 degrees, and twisting angles of 7-13 degrees (Owen & Garg, 1991; Garg et al., 1992; Garg & Owen, 1993). Patient transfers between a wheelchair and shower chair resulted in slightly higher lumbar flexion angles (63-65 degrees), and slightly lower bending and twisting angles (9-11 degrees and 3-9 degrees, respectively; Garg et al., 1991b). Although the lumbar kinematics were reported in these investigations, the risk of low-back injury associated with these lumbar angles remains unclear.

In an attempt to address this issue, Marras et al. (1995) developed a multiple logistic regression model to identify individuals at high risk of occupationally-related low back disorder based on trunk motion and workplace factors. Larger values of lateral bending velocity, twisting velocity, sagittal angle, load moment, and lifting frequency were found to be predictive of high risk of low-back disorder. Based on this model, the average 1.5 degree increase in peak flexion angle of the rear transferor under constrained conditions would increase the odds of being at high risk of injury by a factor of 1.04. This suggests that the spatial constraints imposed by adjacent rows and aisles of seats and by the overhead bins have little adverse effect on the risk of back injury to the rear transferor. A kinetic analysis is needed to confirm this, however, as the presence of the spatial constraints could affect the load distribution between the front and rear transferors.

For the front transferor, the average 8.5 degree increase in peak flexion angle under constrained conditions would increase the odds of being at high risk of low-back injury by a factor of 1.24. We further estimate that the average 10 cm increase in moment arm would increase the load moment by 18.4 Nm and increase the odds of being at high risk of injury by a factor of 1.87 during transfers performed using our 78 kg dummy. Hence, the effects of the spatial constraints appear to markedly increase the risk of low-back injury to the front transferor.

Although the model of Marras et al. (1995) provides preliminary insight into the odds of low-back injury during onboard transfers, these results should be interpreted with caution due to the nature of the lifting tasks involved in the current study. The Marras model was developed based on one-person, highly repetitive lifting tasks, using relatively light loads, not the two-person, single lifts of substantially greater loads commonly associated with dependent transfers. Although the spine may be more resistant to injury during single lifts than during highly repetitive lifting tasks, it may be less able to tolerate motions which result in increased lumbar angles and velocities while supporting greater loads, particularly lifts which require asymmetric trunk motion (Andersson, 1981; Kyserling et al., 1988). Thus, the relationship between trunk kinematics, load moment, and injury risk may differ between the former and latter lifting tasks. Evidence also suggests that there is a difference in the relationship between body dynamics and spinal loading for a one-person versus a two-person lifting task (Marras et al., 1999b).

Nevertheless, it is reasonable that the general relationship between greater peak flexion angle, or a greater load moment, and an increased risk of low-back injury still holds true for the lifting task investigated in the current study. Given this relationship, and the workplace and trunk motion factors found by Marras et al. (1995) to result in high risk group membership, persons performing dependent transfers on board aircraft are at an increased risk of low-back disorders. The estimated load transferred by the rear transferor in the current study was 5 times greater than the load determined by Marras et al. (1995) to predict high risk group membership. Similarly, the estimated maximum load moment and maximum flexion angle in the current study was found to be 4.5 and 3 times greater, respectively, than that found by Marras et al. (1995) to result in high risk group membership.

The difference in the effect of the spatial constraints on the peak kinematics of the front and rear transferors can be explained by the nature (location) of the constraints. Due to the high seatback height of the aircraft seat, the rear transferor must be in a position directly behind the seat in order to lift over it. Placing another seat directly to the rear does not alter his or her transfer method in any way. This suggests that the increase in trunk flexion in the constrained condition is a direct result of the constraints imposed by the adjacent rows and aisles of seats on the front transferor.

In the constrained condition, the front transferor has a seat located directly to his or her side. This prevents him or her from establishing a wide base of support and shifting his or her center of mass laterally during the transfer. As a result, instead of being able to maintain an optimal lifting position, he or she must reach around the seat. This requires a greater degree of lumbar flexion, lateral bending, twisting, and reach (i.e. moment arm) to transfer the transferee to or from the seat, potentially increasing the risk of low-back injury. This compromised lifting position may also prevent the front transferor from generating as much lifting force as in the unconstrained condition. This decrease in force could potentially result in the transferee being raised to a lesser height during the transfer, requiring the rear transferor to assume a position of greater flexion in reaching over the seatback.

Transfers of the large transferee resulted in larger lateral bending angles, lateral bending and twisting velocities, and load moment arms in the rear transferor. Of greatest significance is the increase in load moment arm, resulting in an odds ratio of high risk group membership of 1.88 (Marras et al., 1995) from this single factor alone. Increases in flexion angle during transfers performed with the smaller transferee are likely due to the height of the aircraft seatback and the stature of the transferee. The shorter torso of the small transferee requires the rear transferor to reach farther over the seat while raising or lowering the transferee from this position. Changes in lumbar kinematics of the front transferor during transfers of the large transferee were relatively small and resulted in increased lateral bending angle, and flexion and lateral bending velocity. Given the relatively light weight of the transferees used in the present study, it is reasonable to assume that these effects may be magnified using transferees more representative of the average traveler with disabilities.

The only effect that the direction of transfer had on the kinematic variables of the rear transferor was an increase in peak lumbar flexion velocity during outboard transfers. The tall back of the aircraft seat requires that rear transferors lift up and over the seat during outboard transfers, likely causing them to lower the transferee using only the upper back. During inboard transfers, they are able to maintain a more upright posture throughout the duration of the lift while using their legs to lower the transferee. This may be responsible for the 14 deg/s increase in flexion velocity during the outboard transfers. Of note, large flexion velocities have previously been found to be predictive of high risk of low-back disorder, with an odds ratio of 3.33 (Marras et al., 1993).

The body height of the rear transferor appears to play a major role in the risk of injury to the rear transferor, with shorter transferors experiencing greater lumbar twisting and greater lateral bending and twisting velocities. Although the current study showed no increase in lifting asymmetry (lateral bending and twisting) due to constraint or transfer direction for the rear transferor, differences in trunk bending and twisting of individual transferors may have been masked by the different lifting strategies adopted in order to compensate for height differences. Taller transferors were able to lift the transferee over the seatback, maintaining a more sagittally symmetric lift. Shorter transferors often were required to reach around in front of the seat, as opposed to over the seat, in order to complete the transfer. This strategy resulted in a more asymmetric lifting motion, potentially increasing the amount of lateral bending and twisting.

Although the current study found an influence of transferor height on known kinematic risk factors, two factors not taken into account that may play a role in these findings are those of sex and strength capacity. Previous work by Marras et al. (2002) indicates that women may be at a greater risk of injury than men during specific lifting tasks, particularly during a whole-body dynamic lifting task as performed in the current study. Limited strength capacity has also been found to alter lifting strategies, particularly when the load is lifted to the side (Lee et al., 1990), such as during transfers in which the transferor must reach around or in front of a seat due to either height or constraint limitations. More research is needed to determine the influence of each of these factors on the risk factors for low back injuries during dependent transfers on an aircraft.

Finally, it should be noted that the present results were obtained using an 'ideal' transferee. The use of an anthropometric dummy allowed for more consistent transferee behavior between trials and sessions than would have been achieved using an actual person. However, the rigid nature of the dummy may also have allowed the transfers to be performed more easily than for individuals of similar mass with limited muscular control. In addition, to ensure subject safety, the current study was limited to a maximum dummy size equal to that of a 50th percentile male. The results obtained using a transferee of this size may underestimate the potential risk involved to a transferor while transferring travelers of greater weight and stature.

The results of this study suggest that the spatial constraints imposed by the interior of an aircraft significantly influence the risk of low-back injury to the transferors assisting the traveler. For the front transferor, this increased risk appears to arise from the need to reach around the row of seats in front of the traveler's. It is also possible that the kinematic changes observed in the front transferor could alter the amount of load that is carried by the rear transferor. This additional load, combined with a greater moment arm than that observed during transfers performed in the front transfer position, has the potential to place the rear transferor at a greater risk of injury. Additional analyses are needed to confirm whether this is the case. Rear transferors of short stature also appear to be at greater risk of injury due to the height of the aircraft seatback. These results suggest that the spatial constraints imposed by the interior of an aircraft, and the weight of the traveler being transferred, both contribute to the risk of injury to transferors during dependent transfers performed on an aircraft.

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Table 2.1. Effect of constraint on peak lumbar angles, angular velocities, and load moment arm length of the front transferor (Mean ± SD).

	Constraint Condition	
	Constrained	Unconstrained
Angle (°)		
Flexion *	63.4 ± 12.7	55.8 ± 15.4
Lateral Bending †‡	10.0 ± 3.4	6.6 ± 2.0
Twisting *	4.9 ± 2.3	3.8 ± 1.5
Velocity (°/sec)		
Flexion *‡	20.7 ± 7.3	19.8 ± 6.7
Extension *	16.5 ± 6.6	22.0 ± 10.0
Lateral Bending	21.2 ± 6.8	19.9 ± 10.5
Twisting	13.6 ± 3.9	13.6 ± 5.6
Moment Arm (cm) †§	53.4 ± 5.5	42.0 ± 7.1

* $p < 0.005$; † $p < 0.05$
‡ constraint x direction interaction (Table 2.2; Table 2.6)
§ size x constraint interaction (Table 2.3)

Table 2.2. Constraint by direction interaction effect on peak lateral bending angle (deg) of the front transferor (Mean ± SD).

	Transfer Direction	
	Inboard	Outboard
Constraint		
Constrained	9.1 ± 4.6*	10.9 ± 4.7†
Unconstrained	6.6 ± 2.9*	6.6 ± 2.1†

* $p = 0.001$; † $p < 0.001$ for the comparison between the two conditions indicated.

Table 2.3. Size by constraint interaction effect on peak load moment arm (cm) of the front transferor (Mean ± SD).

	Constraint Condition	
	Constrained	Unconstrained
Size		
Small	54.8 ± 5.5*‡	41.9 ± 7.2*
Large	52.1 ± 6.6†‡	42.1 ± 7.4†

*,† $p < 0.001$; ‡ $p = 0.004$ for the comparison between the three conditions indicated.

Table 2.4. Effect of transfer direction on peak lumbar angles, angular velocities, and load moment arm length of the front transferor (Mean ± SD).

	Transfer Direction	
	Inboard	Outboard
Angle (°)		
Flexion *§	60.6 ± 12.7	58.6 ± 15.4
Lateral Bending *‡	7.8 ± 3.2	8.8 ± 2.9
Twisting	4.4 ± 1.8	4.4 ± 1.9
Velocity (°/sec)		
Flexion *‡	18.9 ± 6.4	21.6 ± 7.4
Extension	19.7 ± 7.9	18.7 ± 8.4
Lateral Bending	21.0 ± 9.0	20.1 ± 8.0
Twisting	13.9 ± 4.0	13.3 ± 5.0
Moment Arm (cm)	47.7 ± 6.3	47.7 ± 5.8

* $p < 0.05$
‡ constraint x direction interaction (Table 2.2; Table 2.6)
§ size x direction interaction (Table 2.5)

Table 2.5. Size by direction interaction effect on peak lumbar flexion angle (deg) of the front transferor (Mean ± SD).

	Transfer Direction	
	Inboard	Outboard
Size		
Small	61.8 ± 13.6*	58.2 ± 14.2*
Large	59.5 ± 14.2	58.9 ± 14.4

* $p < 0.001$ for the comparison between the conditions indicated.

Table 2.6. Constraint by direction interaction effect on peak lumbar flexion velocity (deg/sec) of the front transferor (Mean ± SD).

	Transfer Direction	
	Inboard	Outboard
Constraint		
Constrained	18.3 ± 7.7*	23.2 ± 10.0*
Unconstrained	19.5 ± 8.2	20.1 ± 6.5

* $p = 0.010$ for the comparison between the conditions indicated.

Table 2.7. Effect of transferee size on peak lumbar angles, angular velocities, and load moment arm length of the front transferor (Mean ± SD).

	Size	
	Small	Large
Angle (°)		
Flexion *‡	60.0 ± 15.4	59.2 ± 15.4
Lateral Bending †	7.9 ± 2.8	8.7 ± 2.8
Twisting	4.3 ± 1.6	4.5 ± 1.9
Velocity (°/sec)		
Flexion †	19.2 ± 6.6	21.3 ± 6.4
Extension	19.8 ± 8.6	18.6 ± 7.0
Lateral Bending *	18.8 ± 6.7	22.4 ± 9.9
Twisting	13.0 ± 4.2	14.2 ± 4.8
Moment Arm (cm) †§	48.4 ± 5.8	47.1 ± 6.3

* $p < 0.005$; † $p < 0.05$
‡ size x direction interaction (Table 2.5)
§ size x constraint interaction (Table 2.3)

Table 2.8. Effect of constraint on peak lumbar angles, angular velocities, and load moment arm length of the rear transferor (Mean ± SD).

	Constraint Condition	
	Constrained	Unconstrained
Angle (°)		
Flexion *	58.6 ± 9.9	57.1 ± 10.1
Lateral Bending	8.0 ± 2.5	7.7 ± 2.9
Twisting	6.8 ± 1.9	6.6 ± 2.1
Velocity (°/sec)		
Flexion	26.3 ± 7.6	27.0 ± 7.1
Extension	25.2 ± 7.5	27.7 ± 8.8
Lateral Bending	19.6 ± 4.7	20.3 ± 6.7
Twisting	16.3 ± 7.0	16.7 ± 6.6
Moment Arm (cm)	63.9 ± 5.3	65.0 ± 5.7

*$p < 0.05$

Table 2.9. Effect of direction of transfer on peak lumbar angles, angular velocities, and load moment arm length of the rear transferor (Mean ± SD).

	Transfer Direction	
	Inboard	Outboard
Angle (°)		
Flexion	58.1 ± 10.1	57.5 ± 9.8
Lateral Bending	7.3 ± 2.8	8.4 ± 3.0
Twisting	6.9 ± 2.3	6.6 ± 1.9
Velocity (°/sec)		
Flexion *	19.8 ± 6.2	33.5 ± 9.7
Extension	27.7 ± 9.5	25.1 ± 8.2
Lateral Bending	20.1 ± 6.2	19.8 ± 5.5
Twisting	15.6 ± 6.3	17.3 ± 7.3
Moment Arm (cm)	64.9 ± 6.1	64.0 ± 5.0

* $p < 0.001$

Table 2.10. Effect of transferee size on peak lumbar angles, angular velocities, and load moment arm length of the rear transferor (Mean ± SD).

	Size	
	Small	Large
Angle (°)		
Flexion *	60.4 ± 9.7	55.3 ± 10.1
Lateral Bending *	7.4 ± 2.7	8.3 ± 2.4
Twisting	6.7 ± 1.9	6.7 ± 1.9
Velocity (°/sec)		
Flexion	25.8 ± 6.5	27.5 ± 7.7
Extension	27.6 ± 9.1	25.1 ± 7.8
Lateral Bending †	18.5 ± 5.0	21.4 ± 6.1
Twisting *	15.3 ± 6.3	17.7 ± 6.6
Moment Arm (cm) *	63.2 ± 5.4	65.7 ± 6.0

* $p < 0.005$; † $p < 0.05$

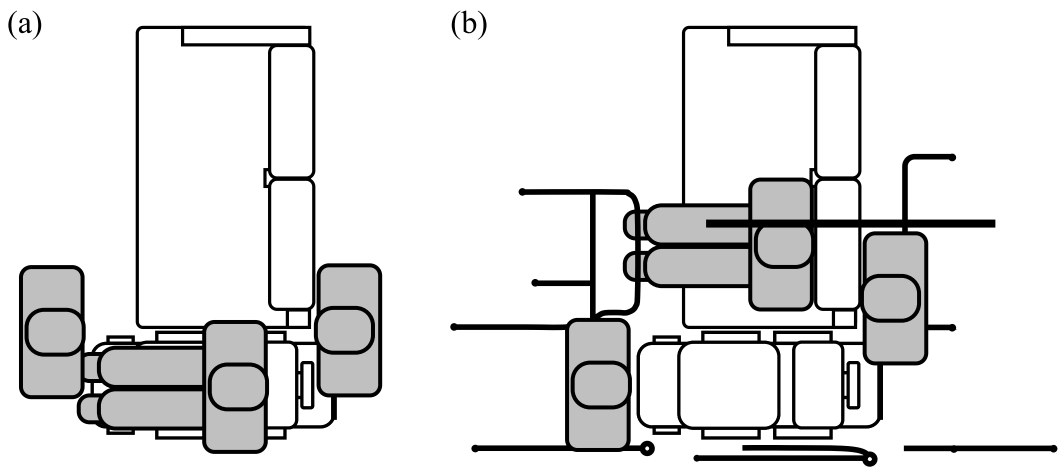

Figure 2.1. Diagram of the experimental set-up showing the configuration at the start of a trial for (a) an outboard dependent transfer in the unconstrained case, and (b) an inboard dependent transfer in the constrained case.

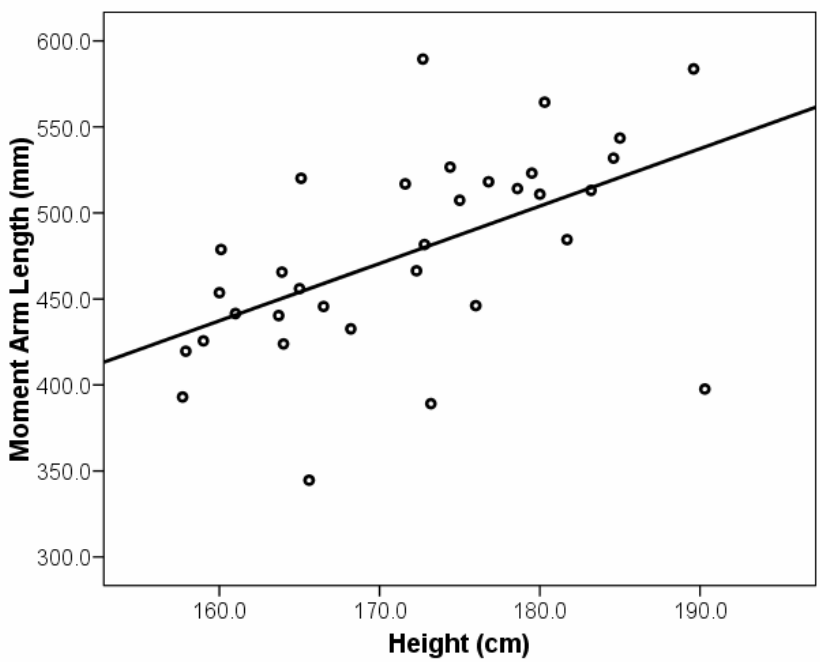

Figure 2.2. Correlation between front transferor body height and peak load moment arm length ($r^2 = 0.29$).

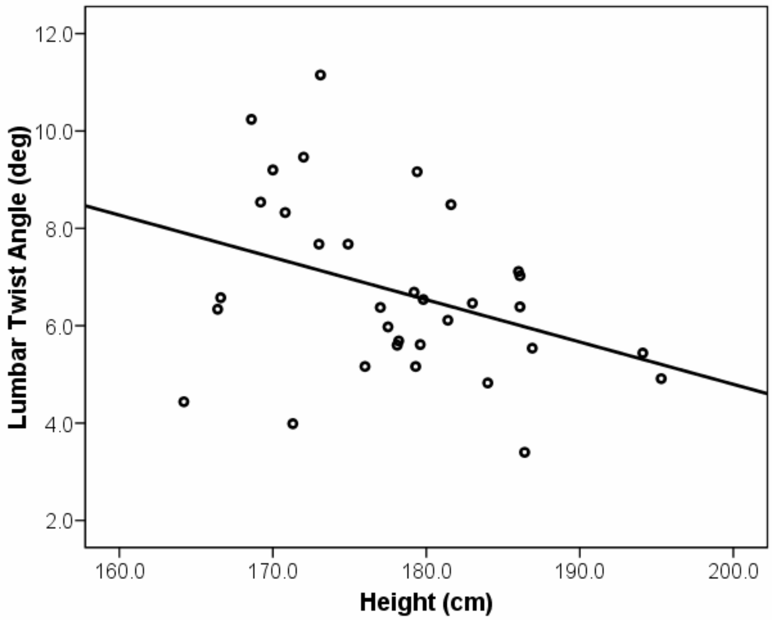

Figure 2.3. Correlation between rear transferor body height and peak lumbar twist angle ($r^2 = 0.13$).

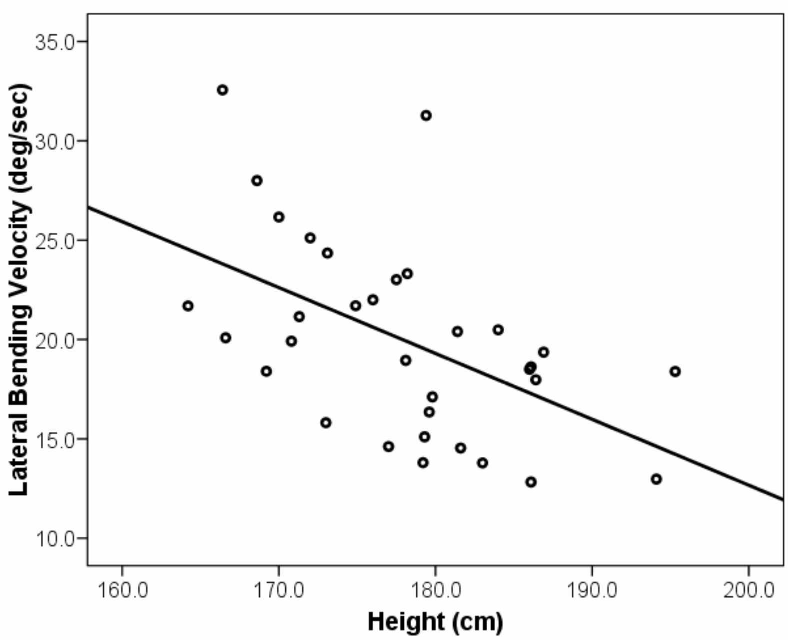

Figure 2.4. Correlation between rear transferor body height and peak lumbar lateral bending velocity ($r^2 = 0.27$).

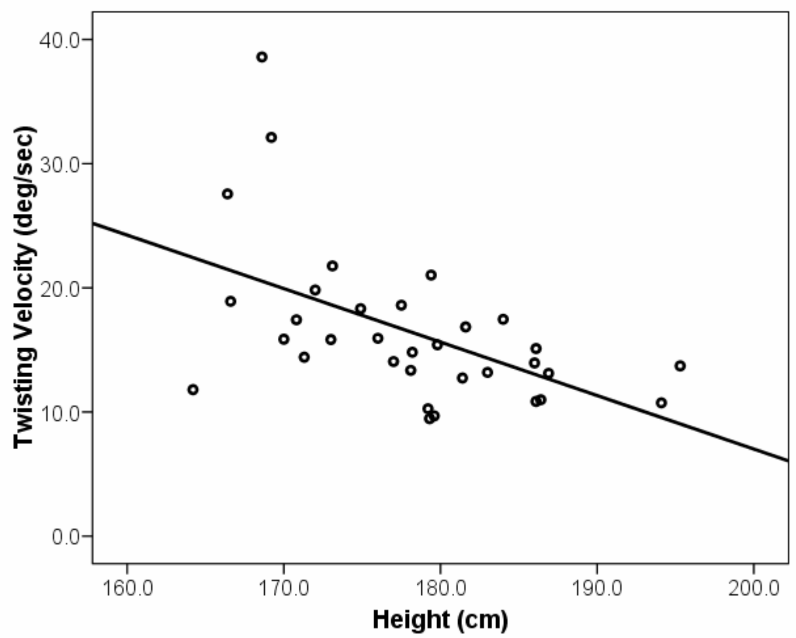

Figure 2.5. Correlation between rear transferor body height and peak lumbar twisting velocity ($r^2 = 0.28$).

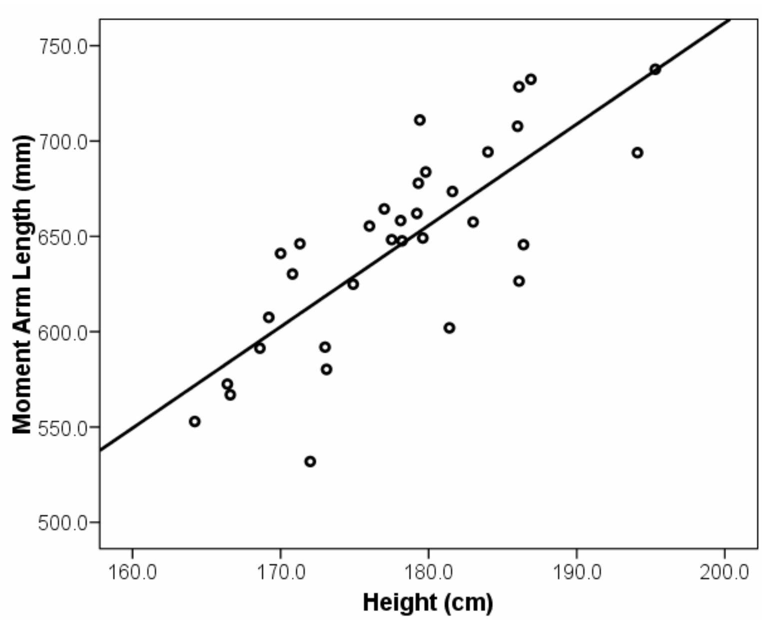

Figure 2.6. Correlation between rear transferor body height and peak load moment arm length ($r^2 = 0.61$).

LUMBAR KINETICS DURING DEPENDENT TRANSFERS ON AN AIRCRAFT

Brian K. Higginson and Michael J. Pavol

3.1 INTRODUCTION

Air travel by people with disabilities poses many challenges to both airport personnel and travelers. From the time they arrive at the terminal to the time they are seated on the plane, travelers are required to undergo multiple transfers to and from various mobility devices. Due to the narrow aisles found on board aircraft, the traveler is unable to use his or her personal mobility aid for boarding. Upon arrival at the boarding gate, the traveler must be transferred to an aisle chair that is narrow enough to fit down the aisle of the aircraft. Once on board the aircraft, they must once again be transferred from the aisle chair to their assigned seat. These dependent transfers on board an aircraft are typically performed using a two-person “through-arm” technique (Pelosi & Gleeson, 1988), potentially placing the transferors at high risk of a disabling low back injury.

Although statistics on the incidence of low back disorders related to the transfers of travelers with disabilities on board an aircraft are not readily available, similar patient handling tasks in clinical and assisted care settings have been found to be associated with high risk of low back disorders (Winkelmolen et al., 1994; Ulin et al., 1997; Marras et al., 1999; Daynard et al., 2001; Skotte, 2001; Schibye, et al., 2003). The high rate of low back disorders in patient handlers can be attributed to factors that may otherwise be prevented, or controlled for, during lifting tasks commonly found in industry. Numerous studies have shown an increase in lumbar compression forces and moments as the weight of the external load being lifted is increased (Freivalds, et al., 1984; Jager & Luttman, 1989; Schipplein et al., 1990; Garg & Owen, 1992; Granata &

Marras, 1995; Ulin et al., 1997; Granata et al., 1999). Unfortunately, during patient handling tasks, the load being transferred cannot be modified as in manual materials handling tasks, resulting in greater risk of injury to the transferors. Patient transfers also tend to require transferors to assume a more awkward lifting position than found among lifting tasks in industry, resulting in a more asymmetrical lifting posture.

Patient transfer tasks most similar to those performed on board an aircraft (transfers between wheelchair and toilet, and wheelchair and bed) have been shown to result in greater compressive forces and moments at the lumbar spine than in any other patient handling task (Garg et al., 1992). However, the dependent transfer of a traveler with disabilities on board an aircraft poses unique challenges to the transferors that may further increase the risk of injury. The interior of an aircraft imposes rigid spatial constraints that are potentially the most extreme found in any patient handling environment. Constraints include the adjacent rows and aisles of seats, the overhead bins, and the high seatback of the aircraft seat itself. Due to these constraints, the transferors must adopt a transfer technique unlike those most commonly analyzed in the patient handling literature. Most investigations of patient handling tasks using a two-person transfer method have been conducted for techniques in which both transferors lift from the front of the person being transferred (Garg et al., 1991; Owen & Garg, 1991; Garg et al., 1992; Garg & Owen, 1993; Marras et al., 1999; Daynard et al., 2001). Due to the constraints imposed by the interior of the aircraft, this transfer method is not possible on board an aircraft; one transferor must lift from the front, while the other lifts from the rear. The influence of this transfer method on risk factors associated with low-back disorders has yet to be established.

The use of this transfer technique likely results in a larger proportion of transferee weight being borne by the rear transferor than would be expected using the front transfer technique studied by previous investigators, potentially increasing the risk of injury to the rear transferor. It is also unclear how the constraints imposed by the interior of the aircraft influence this risk. It is possible that, given the presence of the constraints, the use of this transfer technique may result in lifts being performed faster and may require more changes in direction of trunk motion during the transfer, accommodations in lifting technique that have been previously found to accompany tasks involving handling large body weights, and shown to increase the risk of low back disorders (Gagnon et al., 1987; Busek et al., 1988; Gagnon & Smyth, 1992; Lavender et al., 1999).

To date, the effects of these constraints on kinetic risk factors associated with low-back disorders have not been determined in persons performing dependent transfers on an aircraft. Nor is it known whether these risk factors differ as a function of transfer direction (i.e. wheelchair-to-seat or seat-to-wheelchair), or size of the individual being transferred. The purpose of the current study was to determine the extent to which these factors influence kinetic risk factors of rear transferors during two-person dependent transfers on board an aircraft. By understanding these effects, appropriate means of reducing the risk of injury may be developed.

3.2 METHODS

Study participants included 33 pairs of men (n = 42) and women (n = 24) between the ages of 18 and 40 years old, each pair consisting of a front transferor and a rear transferor. The mean ± SD age, mass, and height were 24.1 ± 6.4 years, 68.8 ± 13.3 kg, and 172.0 ± 9.4 cm, respectively, for the front transferors, and 23.6 ± 4.2 years, 78.0 ± 13.7 kg, and 178.0 ± 7.7 cm for the rear transferors. Health history questionnaires were administered to screen participants for contraindications to the level of physical exertion required to perform a dependent transfer. Eligible subjects consisted of healthy individuals who were free from pre-existing back disorders or injuries and who had regularly been resistance training or the equivalent over the preceding month Subjects were not required to have previous experience performing dependent transfers. Institutional Review Board approval was obtained and subjects provided written informed consent prior to participation.

Subjects worked in pairs to transfer an anthropometric dummy between a wheelchair and an airplane seat. The two-person, dependent transfers were performed in a laboratory simulation of an aircraft interior. Transfers were performed using the "through arm" transfer technique (Pelosi and Gleeson, 1988). In this technique, the rear transferor reaches beneath the shoulders of the transferee, grasps the forearms, and secures them across the chest. The front transferor reaches under the thighs and grasps the transferee just above the knees. The transferors lift and shift the transferee sideways synchronously upon a verbal three-count initiated by the rear transferor.

Transfer position (front or rear) was self-selected by the subjects, and was maintained for the duration of testing.

A standard economy-class airplane seat (Boeing, Seattle, WA) was mounted to the floor with the armrest raised. An aircraft aisle wheelchair (AisleMaster 8000, Columbia Medical, Pacific Palisades, CA) was placed adjacent to the aircraft seat and secured in place. A small anthropometric dummy (Simulaids, Saugerties, NY), representative of a 50th percentile woman (mass: 57 kg; height: 165 cm), and large (First Technology Safety Systems, Plymouth, MI) anthropometric dummy, representative of a 50th percentile male (mass: 78 kg; height: 178 cm), served as the transferee during all dependent transfers.

Removable frames, constructed from electrical conduit, were used to simulate the spatial constraints imposed by the aircraft interior (Figure 3.1). These constraints included the surrounding rows and aisles of seats and the overhead bins, and were placed at distances approximating those found on a typical single-aisle aircraft. Aisle width was set at 43.2 cm, row pitch (distance from the front of one seat to the front of the next seat) was 78.4 cm, and overhead bins were located at a height of 162.5 cm above the floor and 30.5 cm outboard from the inboard (i.e. aisle) edge of the seat.

Subjects were instructed on proper transfer technique and, after a warm-up, were allowed two practice transfers prior to testing in order to familiarize themselves with the required transfer task. The subjects then transferred each dummy once for each transfer direction (outboard: wheelchair-to-seat, and inboard: seat-to-wheelchair) under constrained and unconstrained conditions, for a total of eight trials. Trials were blocked by constraint, and within constraint, by dummy size. Blocks, and trials within blocks, were counterbalanced to account for any potential order effects. At least two minutes of rest was provided between trials.

Three-dimensional body segment motions were recorded during each transfer by using an eight-camera motion capture system (Vicon, Lake Forest, CA) to track 74 reflective markers attached to the subjects and the dummy. Ground reaction forces on the feet of the rear transferor were acquired through the use of two force plates (Bertec, Columbus, OH) mounted flush with the floor immediately to the rear of the aircraft seat. The magnitude and center of pressure of the compressive forces between the right thigh of the rear transferor and the aircraft seat were measured utilizing a custom-built force plate mounted to the aircraft seat back (Schafer et al., 2005).

Marker trajectories were sampled at 60 Hz and low-pass filtered using a 4^{th}-order zero-lag Butterworth filter at a cut-off frequency of 8 Hz, as determined by residual analysis of the raw marker position data (Winter, 1990). The recorded marker locations, along with anthropometric measurements and the marker positions during a trial of quiet standing, were used to determine the three-dimensional orientations of the feet, legs, thighs, pelvis, and trunk during each trial. The spatial positions of the joint centers at the ankles, knees, hips, and a lumbar joint located at the L_3/L_4 level were similarly determined.

Resultant lumbar joint forces and moments were calculated based on a seven-link rigid segment model using a bottom-up, three-dimensional inverse dynamics approach (BodyBuilder, Vicon, Lake Forest, CA). Modeled body segments included the feet, legs, thighs, and pelvis. Body segment masses were calculated from measured body dimensions using regression equations from Pavol et al. (2002), with center of mass locations and mass moments of inertia obtained from de Leva (1996).

Reaction forces from the three force plates were sampled at 600 Hz and used to solve the equations of motion needed to determine forces and moments at the lumbar joint center. Given the position of the rear transferor relative to the aisle chair and aircraft seat, it was assumed that any contact made would occur between the seatback and the right thigh of the transferor. Therefore, contact forces measured at the seatback were applied to this segment of the kinetic model. The equations of motion were solved in a distal-to-proximal order, starting at the feet. Lumbar compressive forces were expressed along the longitudinal axis of the pelvis. Lumbar moments about the three principal axes of rotation were expressed based on a body-fixed rotation sequence of flexion-extension, bending, and twisting of the trunk relative to the pelvis. These resultant joint forces and moments were then normalized by body weight (BW), and body weight and height (BWxHT), respectively.

Dependent measures consisted of the peak resultant lumbar compressive force and the peak lumbar moments about the three principal axes of motion. Right and left bending and twisting were considered to be equivalent (i.e. the absolute value was used) in the determination of the corresponding peak bending and twisting moments. Three-way (2 x 2 x 2) repeated measures analyses of variance were used to test the effects of constraint, transferee size, and the direction of transfer on the dependent measures. Effects were considered significant at the .05 alpha level. Calculation of descriptive statistics, tests of model assumptions, and statistical analyses were performed using SPSS version 13.0 (SPSS, Chicago, IL).

3.3 RESULTS

For all conditions, peak extension moments were consistently greater than either the peak bending or twisting moments (~19.5 %BWxHT for extension vs. 4.5 and 3.4 %BWxHT for bending and twisting, respectively). Of the three factors investigated, transferee size had the most notable influence on peak lumbar kinetics of the rear transferor during the dependent transfers (Table 3.1). Transfers of the large transferee resulted in greater peak lumbar compressive forces, as well as greater peak extension, lateral bending, and twisting moments (all $p < 0.001$). On average, the peak compressive force was 14.0 %BW greater while transferring the large dummy as compared to the small dummy. Smaller differences in lumbar moments were found, with transfers of the large dummy resulting in peak extension, bending, and twisting moments that were 3.2, 1.3, and 0.9 %BWxHT larger, respectively, than for transfers of the small dummy.

The constraints imposed by the interior of the aircraft had relatively little effect on peak lumbar kinetics of the rear transferor, resulting only in a slightly smaller (0.5% BWxHT) peak bending moment during the transfers in which the constraints were present ($p < 0.01$; Table 3.2). The direction of transfer appeared to have no influence on peak lumbar moments (Table 3.3), whereas peak lumbar compressive force was approximately 4 %BW greater during outboard transfers ($p < 0.001$).

3.4 DISCUSSION

Due to the nature of dependent transfers on board an aircraft, the persons performing the transfer are at risk of disabling low back injury. Assuming that a larger proportion of transferee weight is borne by the rear transferor during these transfers, transferors in this position may be at an increased risk of injury. The purpose of the current study was to determine the extent to which ergonomic factors such as the spatial constraints imposed by the aircraft interior, direction of transfer, and the weight of the transferee influence selected kinetic risk factors associated with low-back disorders among rear transferors performing dependent transfers on board an aircraft.

Lumbar kinetics were relatively unaffected by the presence of surrounding rows and aisles of aircraft seats, and of overhead bins. In one respect, these results are not surprising given that the addition of these constraints does not require the rear transferor to significantly modify his or her lifting technique in order to perform the transfer. Interestingly, lateral bending moments decreased slightly during transfers in which these constraints were present. This observation may potentially be explained by the limited ability of the front transferor to shift his or her pelvis laterally during the transfer, given the presence of the row of seats in front of that of the traveler. This limited mobility may prevent the front transferor from contributing equally in the transfer of the load, requiring the rear transferor to carry a larger percentage of the transferee weight. In order to adapt to this increased loading requirement, the rear transferor may be required to make a more deliberate transfer of weight from one foot to the other in order to comfortably transfer the additional weight, thereby decreasing the amount of lateral bending moment observed during the constrained condition.

During unconstrained transfers the front transferor is able to transfer his or her weight from one foot to the other, affording a more ideal lifting position to support a greater portion of the load throughout the transfer. This may allow the rear transferor to adopt a more neutral, stationary position between the aisle chair and aircraft seat during the transfer, a position which results in more lateral bending at the beginning and end of the transfer.

Of the kinetic variables investigated, resultant lumbar compressive force was the only variable influenced by the direction of transfer, with greater compressive forces experienced by the rear transferor during outboard transfers. This finding may be attributable to the lower height of the aisle chair seatback compared to that of the aircraft seat, which likely alters the posture assumed by the transferor upon initiation of the lift. The lower seatback height of the aisle chair facilitates a greater degree of trunk flexion at the initiation of the lift than does the high seatback height of the aircraft seat. This increased trunk flexion may lead to greater trunk acceleration during extension at the onset of the lift and greater changes in the direction of trunk motion during the lift, both of which have been found to result in increased lumbar loading. (Susnik & Gazvoda, 1986; Gagnon et al., 1987; Marras & Mirka, 1991; Jager & Luttman, 1999; Duan et al., 2001). Garg et al. (1991) found no difference in erector spinae muscle force or lumbar compression force between transfer directions for a similar transfer task (wheelchair to shower chair vs. shower chair to wheelchair), supporting the idea that the height of the aircraft seat back may potentially be responsible for the difference found in resultant lumbar compressive forces between inboard and outboard transfers in the current study. For taller transferors able to reach over the seatback in order to perform the transfer, the seatback may have provided additional support while initiating the lift during inboard transfers, thereby reducing the compressive force acting on the lumbar joint in that direction.

The results of this study indicate that, of the factors investigated, transferee size has the greatest influence on known risk factors for low back disorder. Transfers involving the large (78 kg) transferee resulted in a greater resultant lumbar compressive force, as well as greater peak extension, lateral bending, and twisting moments about the lumbar joint center. These results are to be expected, as a simple static analysis of the lifting task would confirm that resultant lumbar compressive forces would increase in proportion to the magnitude of the applied external load above the point of interest. Similarly, at any given lumbar angle, an increase in external load will lead to a corresponding increase in the load moment. Greater estimated lumbar compression forces and load moments at the lumbar joint center during lifting-specific tasks as a result of lifting greater external loads have also been found by others (Gagnon et al., 1985, 1987; Buseck et al., 1988; Garg et al., 1991; Dolen et al., 1994; Ulin et al., 1997; Lavender et al., 1999), although actual differences in estimated values vary depending on such factors as transferor stature, the magnitude of externally applied loads, and the method in which the compressive forces are estimated.

One factor that needs to be considered in determining the effects of transferee size on compressive force is the weight of the transferor. A transferee of a given weight will represent a greater load, in proportion to body weight, for a smaller transferor. Furthermore, there is evidence to suggest that vertebral cross-sectional area, bone density, and compressive strength are greater in heavier individuals (Gilsanz et al., 1994; Duan et al., 2001a, 2001b; Riggs et al., 2004; Bouxsein et al., 2006) enabling then to withstand greater absolute loads. These factors have not been considered in other studies, which report loading in Newtons. For this reason, the lumbar loads analyzed in the current study were normalized by body weight for compressive force estimates, and body weight and height for moment estimates. Normalization of the current data indicates that smaller transferors are at a greater risk of injury while performing these transfers (Figure 3.2). These results agree with others that have shown smaller individuals experience greater mechanical stress within the vertebral bodies under similar loading conditions (Gilsanz et al., 1994). For comparison purposes, the same data are presented without normalization (Figure 3.3). Expressed in this manner, it would appear that larger people are at a higher risk of injury, as the resultant compressive forces are greater. However, these results may be misleading, given what is known about the relationship between stature and the properties of the vertebrae that enable them to resist externally applied loads.

In part, the association observed between lumbar compressive forces and the weight of the transferors may also be a function of the lifting strategy adopted by the transferors due to the high seatback of the aircraft seat. It is possible for larger transferors, capable of lifting over the seatback, to use the seatback for support during the transfers, whereas smaller transferors often adopt a lifting strategy which requires them to reach around the side of the seat to complete the transfer. This strategy prevents the use of the seat back as a support, potentially increasing the resultant lumbar compressive forces for smaller transferors.

The method used to calculate lumbar compressive forces in the current study may result in an underestimation of the actual differences in lumbar loading between conditions. External forces were used to calculate net joint reaction forces acting on the lower lumbar spine. This method of estimating lumbar loading fails to take into account the internal loading created by the muscles providing trunk motion and stabilization. Neglecting the contribution of internal loading has been found to underestimate the total compression forces experienced by the motion segments of the spine by as much as 45% (Granata & Marras, 1995). In more upright postures, the effects of internal loading are minimal. In more flexed positions, the muscles responsible for providing the extension moment and the muscles activated to stabilize the vertebral motion segments in this position contribute to an increased internal loading.

Upon visual inspection of the raw video data, it was apparent that individuals adopt different lifting strategies in order to complete the transfer task. It is reasonable to expect that these differences in transfer techniques could potentially lead to different lumbar loading patterns. Therefore, it may be beneficial for future studies to account for different lifting strategies used by the transferors in order to elucidate their influence on lumbar loading. Also, due to the location of the surrounding rows of seats, the constraints found on board an aircraft may result in a greater risk of injury to the front transferor than was found for the rear transferor in the current study. Further studies are needed to determine the influence of theseconstraints on the front transferor, as there may be a greater influence on lumbar loading in this position.

For logistical and inter-trial repeatability reasons, anthropometric dummies were chosen to be used as transferees. It is unclear what effects this may have had on the current results. However, there is evidence to suggest that lifting a manikin results in higher compression and shear forces at L_5/S_1 than does lifting an actual person (Gagnon et al., 1986). Due to concerns for subject safety, the weight of the transferees used in the current study was relatively small. The 78 kg transferee increased the estimated lumbar compression force by 14 %BW over that of the 57 kg transferee. Travelers with disabilities exceeding the 78 kg used in the current study would be expected to present higher lumbar loads than those shown here.

These results suggest that the transfer of larger individuals poses a greater risk of lower lumbar spine injury to the rear transferor during dependent transfers on an aircraft than do either the spatial constraints or the direction of transfer. The potential for increased lumbar loading also appears to be greater during transfers performed by smaller transferors.

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Table 3.1: Peak lumbar compressive forces and moments (Mean ± SD) while transferring the small and large transferees.
BW: body weight; HT: height

	Size	
Variable	Small	Large
Comp. Force (%BW) *	109.2 ± 14.7	123.2 ± 20.9
Moments (%BWxHT)		
Extension *	18.0 ± 2.8	21.2 ± 5.1
Bending*	3.8 ± 1.1	5.1 ± 1.1
Twisting *	2.9 ± 1.1	3.8 ± 1.1

* $p < 0.001$

Table 3.2: Peak lumbar compressive forces and moments (Mean ± SD) during constrained and unconstrained transfers.
BW: body weight; HT: height

	Constraint	
Variable	Constrained	Unconstrained
Comp. Force (%BW)	116.5 ± 18.7	115.8 ± 17.0
Moments (%BWxHT)		
Extension	19.7 ± 4.0	19.4 ± 4.0
Bending *	4.2 ± 1.1	4.7 ±1.1
Twisting	3.4 ± 1.1	3.3 ± 1.1

*$p < 0.01$

Table 3.3: Peak lumbar compressive forces and moments (Mean ± SD) during inboard and outboard transfers.
BW: body weight; HT: height

	Direction	
Variable	Inboard	Outboard
Comp. Force (%BW) *	114.2 ± 18.7	118.1 ± 17.0
Moments (%BWxHT)		
Extension	19.5 ± 4.0	19.6 ± 4.0
Bending	4.5 ± 1.1	4.3 ± 1.1
Twisting	3.4 ± 1.1	3.3 ± 1.1

*$p < 0.001$

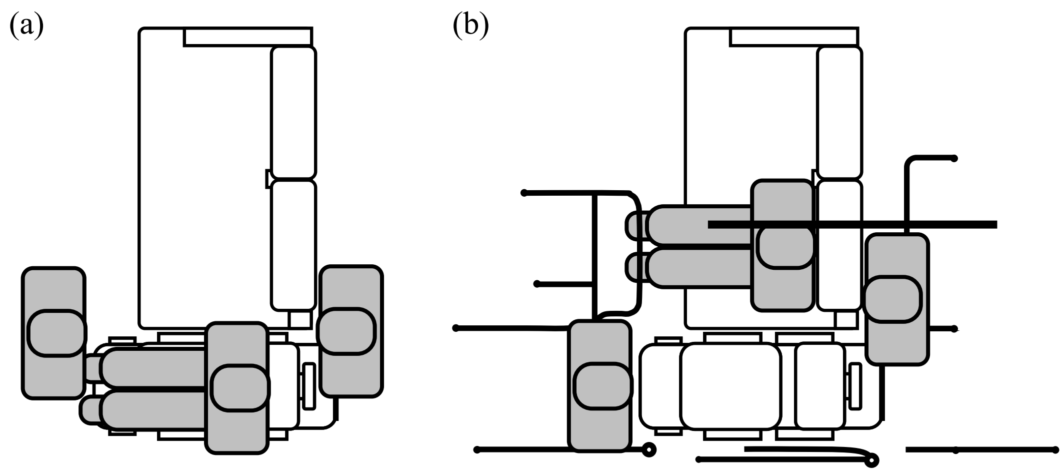

Figure 3.1. Diagram of the experimental set-up showing the configuration at the start of a trial for (a) an outboard dependent transfer in the unconstrained case, and (b) an inboard dependent transfer in the constrained case.

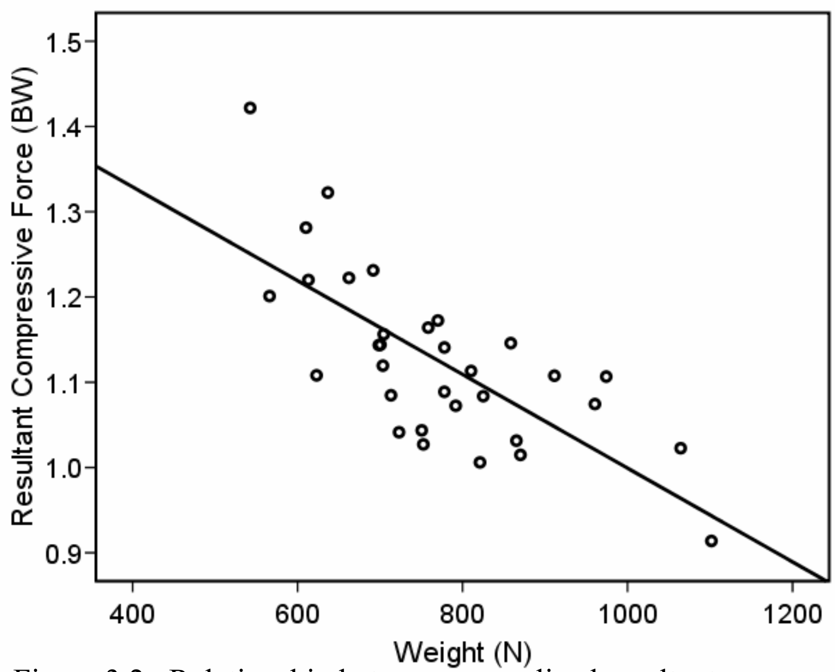

Figure 3.2. Relationship between normalized resultant compressive force and rear transferor weight ($r^2 = 0.53$; $p < 0.001$).
BW= body weight

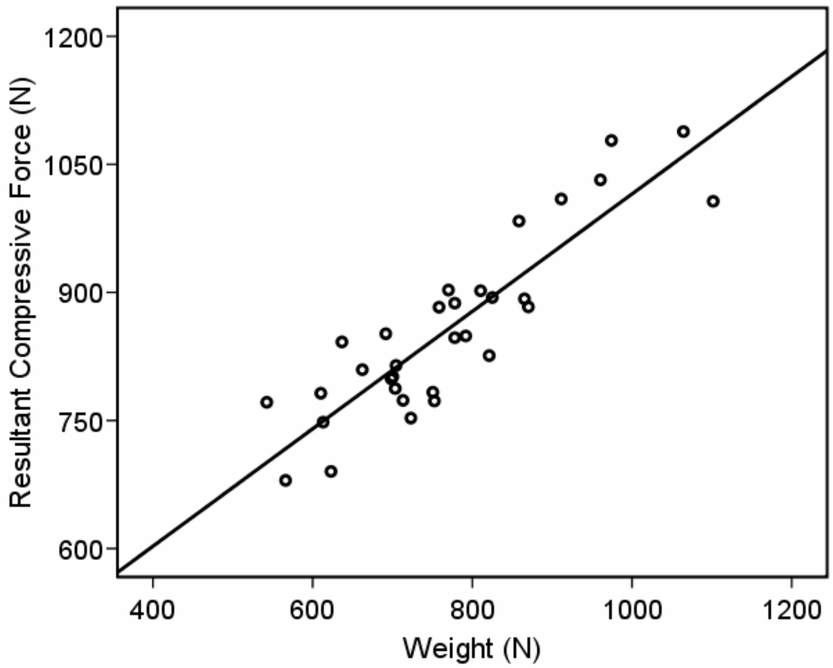

Figure 3.3. Relationship between raw resultant compressive force and rear transferor weight ($r^2 = 0.78$; $p < 0.001$).

PERCEIVED TASK DIFFICULTY AND TRANSFEREE ACCELERATION DURING DEPENDENT TRANSFERS ON AN AIRCRAFT

Brian K. Higginson and Michael J. Pavol

4.1 INTRODUCTION

The ability for people with disabilities to travel by commercial aircraft is important for their full participation in society, but poses many challenges. From their arrival at the airport until they board the aircraft, travelers may be required to perform multiple transfers to and from various mobility aids. Due to the spatial constraints imposed by the interior of the aircraft, travelers with disabilities must be transferred to an aisle chair prior to boarding. The aisle chair is narrower than conventional wheelchairs, allowing it to be pushed down the aisle of the aircraft. Once on board the aircraft, the traveler must once again be transferred, this time from the aisle chair to the aircraft seat. An area of great concern during these repeated transfers is the risk of injury to the traveler, particularly during transfers that take place on board the aircraft, as this presents the most challenging environment in which to safely perform such a transfer. During these dependent transfers, the traveler is reliant on the ability of the transferors to safely execute the task, and could potentially suffer injury if dropped or mishandled.

The risks of musculoskeletal injuries to persons performing dependent transfers in the clinical and assisted care settings have been well documented (Marras et al., 1999; Schibye, et al., 2003; Skotte et al., 2002; Ulin et al., 1997; De Looze et al., 1994). However, very little work has been done to quantify the risk of injury to the individual being transferred. The only studies to date specifically looking at injury risk to the individual being transferred use subjective measures, such as comfort and security ratings as perceived by the transferee during the transfer (Garg et al., 1991a, 1991b; Owen & Garg, 1991). In these studies, the highest discomfort and insecurity ratings were associated with the transfer tasks most similar to those performed on board an aircraft. Patients reported high levels of discomfort (6.7 out of 7) and insecurity (6.1 out of 7) during two-person manual transfers from a toilet to wheelchair (Owen & Garg, 1991). Similar results were found during two-person manual transfers of a patient from a shower chair to wheelchair (6.8 for discomfort, 5.3 for insecurity; Garg et al., 1991b). To date, there has been no study utilizing more objective measures to quantify risk factors related to an increased risk of injury to the individual being transferred. Risk factors resulting in increased odds of sustaining soft tissue damage and lumbar compression injuries during a transfer, such as vertical accelerations, can be objectively measured through the use of instrumentation such as accelerometers.

It is possible that, in patient transfer tasks, the individual being transferred is exposed to a greater risk of injury as the perceived exertion of the individuals performing the transfer increases. One of the most common metrics for quantifying 'effort' or 'exertion' is the Ratings of Perceived Exertion (RPE) scale (Borg, 1970). This 15 point scale (6-20) is used to rate perceptions of exertion over a range encompassing "very, very light" (7) through "very, very hard" (19). Ratings of perceived exertion using this scale have been found to be highly correlated with many physiological variables, including heart rate and oxygen consumption (Borg & Noble, 1972), and have been validated using various modes of work and exercise (Gambarale, 1972; Stamford, 1976; Borg, 1973; Noble et al., 1973; Skinner et al., 1973). More recently, exertion has been used successfully in industrial and clinical care settings to determine the potential for injury during lifting and patient transfer tasks. During dependent transfers performed on board an aircraft, many factors may contribute to an increase in perceived exertion, including the weight of the person being transferred, the spatial constraints imposed by the interior of the aircraft, and the direction in which the transferee is being transferred (into or out of the aircraft seat).

The purpose of this study was to determine the effect of the spatial constraints of the aircraft interior, the transferee size, and the direction of transfer on the vertical acceleration experienced by the transferee. It was also of interest to determine the effect of these factors on the perceived exertion of the individuals performing the transfer and, ultimately, whether these perceptions of task difficulty influence the accelerations experienced by the traveler during the transfer.

4.2 METHODS

Study participants consisted of 33 pairs of men (n = 42) and women (n = 24) between the ages of 18 and 40 years, each pair consisting of a front transferor and a rear transferor. The mean ± SD age, mass, and height were 24.1 ± 6.4 years, 68.8 ± 13.3 kg, and 172.0 ± 9.4 cm, respectively, for the front transferors, and 23.6 ± 4.2 years, 78.0 ± 13.7 kg, and 178.0 ± 7.7 cm for the rear transferors. Health history questionnaires were administered to screen participants for contraindications to the level of physical exertion required to perform a dependent transfer. Eligible subjects consisted of healthy individuals free from pre-existing back disorders or injuries. Subjects were not required to have previous experience performing dependent transfers. Institutional Review Board approval was obtained and subjects provided written informed consent prior to participation.

Subjects worked in pairs to transfer an anthropometric dummy between a wheelchair and an airplane seat. The two-person, dependent transfers were performed in a laboratory simulation of an aircraft interior using the “through arm” transfer technique (Pelosi & Gleeson, 1988) in which the rear transferor reaches beneath the shoulders of the transferee, grasps the forearms, and secures them across the chest. The front transferor reaches under the thighs and grasps the transferee just above the knees. The transfers were performed synchronously using a verbal three-count initiated by the rear transferor. Transfer position (front or rear) was self-selected by the subjects and was maintained for the duration of testing.

A standard economy-class airplane seat (Boeing, Seattle, WA) was mounted to the floor with the armrests raised. An aircraft aisle wheelchair (AisleMaster 8000, Columbia Medical, Pacific Palisades, CA) was placed adjacent to the aircraft seat and secured in place. Either a small anthropometric dummy (Simulaids, Saugerties, NY), representative of a 50th percentile woman (mass: 57 kg; height: 165 cm), or a large (First Technology Safety Systems, Plymouth, MI) anthropometric dummy, representative of a 50th percentile man (mass: 78 kg; height: 178 cm), served as the transferee during all dependent transfers.

Removable frames, constructed from electrical conduit, were used to simulate the spatial constraints imposed by the aircraft interior. These constraints included the surrounding rows and aisles of seats and the overhead bins, and were placed at distances approximating those found on a typical single-aisle aircraft. Aisle width was set at 43.2 cm, row pitch (distance from the front of one seat to the front of the next seat) was 78.4 cm, and overhead bins were located at a height of 162.5 cm above the floor and 30.5 cm outboard from the inboard (i.e. aisle) edge of the seat.

Subjects were instructed on proper transfer technique and, after a warm-up, were allowed two practice transfers prior to testing in order to familiarize themselves with the required transfer task. The subjects then transferred each dummy once for each transfer direction (outboard: wheelchair-to-seat; inboard: seat-to-wheelchair) under constrained and unconstrained conditions, for a total of eight trials. Trials were blocked by constraint, and within constraint, by dummy size. Blocks, and trials within blocks, were counterbalanced to account for any potential order effects. At least two minutes of rest was provided between trials.

Upon completion of each trial, the subjects rated their perceived exertion during the transfer using the 15-point Borg Rating of Perceived Exertion scale (Borg, 1970). Vertical accelerations experienced by the dummies during each transfer were also measured using a high-frequency uniaxial accelerometer (Model 352B10, PCB Piezotronics, Depew, NY) securely attached to each dummy. The accelerometer was mounted vertically at the level of the lower abdomen, providing an estimate of the vertical impact force between the dummy and the seating surface. The accelerometer signal was amplified by a factor of 10 and sampled at 600 Hz. Peak accelerations were determined for each transfer after converting the raw signal to units of gravity (g's) and low-pass filtering at a cut-off frequency of 15 Hz using a fourth-order Butterworth filter.

Three-way (2x2x2) repeated measures analysis of variance (ANOVA) were used to test the effects of dummy size, constraint, and transfer direction on the acceleration and perceived exertion measures. Effects were considered significant at an α of 0.05. Pearson product moment correlations were used to establish relationships between the perceived exertion of the transferors and the peak vertical acceleration experienced by the transferee across all subject pairs and transfer tasks. Calculation of descriptive statistics, tests of model assumptions, and statistical analyses were performed using SPSS version 13.0 (SPSS, Chicago, IL).

4.3 RESULTS

The peak vertical accelerations experienced by the transferee during the dependent transfers were found to be independent of the size of the individual being transferred ($p = 0.261$; Table 4.1), as well as the constraints imposed by the interior of the aircraft ($p = 0.661$; Table 4.2). The direction of transfer influenced the peak vertical acceleration experienced by the traveler ($p = 0.002$), with greater accelerations occurring during inboard transfers (Table 4.3). For all transfer conditions, there was a large between-group variability associated with the vertical acceleration measure.

Ratings of perceived exertion (RPE) increased by 2 points, on average, for transfers performed with the larger transferee (Table 4.1) and during transfers performed in the presence of surrounding constraints (Table 4.2) for both the front and rear transferor ($p < 0.005$). The direction of transfer appeared to have no effect on the perceived exertion of either the front ($p = 0.150$) or the rear ($p = 0.101$) transferor during execution of the transfer task (Table 4.3).

Finally, there was no relationship between the peak vertical acceleration experienced by the transferee and the perceived exertion of either the front ($r = -.065$; $p = 0.331$; Figure 4.1) or the rear ($r = -.091$; $p = 0.176$; Figure 4.2) transferor during the transfer tasks.

4.4 DISCUSSION

The current study was undertaken to objectively quantify the vertical acceleration experienced by travelers with disabilities during dependent transfers performed on board an aircraft, and to elucidate the effects of factors that may influence this risk factor for injury. Factors investigated included the environmental constraints imposed by the interior of the aircraft, the size of the individual being transferred, and the direction in which the transfer was performed. Also of interest was the role these factors played in the perception of task difficulty by the persons performing the transfers, and if differences in perceived exertion resulted in a change in magnitude of the vertical acceleration experienced by the traveler.

Of the factors investigated during dependent transfers onboard an aircraft, transfer direction appeared to be the only factor that had a significant influence on the magnitude of the vertical acceleration experienced by the transferee, with inboard transfers resulting in a 40% increase in vertical acceleration. On average, peak vertical acceleration measures at the levels of the other factors fell between 0.40 and 0.44 g's. The larger peak vertical accelerations associated with inboard transfers may be due to differences in the characteristics of the two seating surfaces. The aircraft seat was well padded, whereas the seat of the aisle chair was firm, with little padding. This lack of cushioning could lead to greater accelerations experienced upon seat impact, potentially increasing the risk of injury to the transferee, particularly if dropped or not otherwise transferred in a controlled manner.

Unlike the peak vertical acceleration experienced by the transferee, the perceived exertions of the front and rear transferors were insensitive to transfer direction, but were significantly larger during transfers involving larger transferees, as well as transfers performed in the presence of constraints. On average, the perceived exertion during these trials increased from 10 (between ‘very light’ and ‘fairly light’) during the unconstrained and small transferee conditions, to 12 (between ‘fairly light’ and ‘somewhat hard’) during the constrained and large transferee conditions. The observed effect of transferee size on perceived exertion was expected, as greater forces are required to lift a heavier load. A similar effect of transferee size on perceived exertion was found by Ulin et al. (1997) while performing similar transfers with 47 kg and 91 kg patients.

Both front and rear transferors perceived transfers performed in the presence of constraints to be more difficult, but this effect was unrelated to the weight of the load being lifted. During transfers performed in the constrained condition, the front transferor is forced to stand in the aisle and reach around the aircraft seat in front of the traveler’s, resulting in greater trunk flexion, lateral bending, and twisting (Higginson et al., 2007). These types of trunk motion have been shown to be correlated with greater ratings of perceived exertion (Winkelmolen, 1994). However, this association between trunk motion and perceived exertion doesn't adequately explain the increase in perceived exertion of the rear transferor, since the lifting motion of the rear transferor is relatively unaffected by the constraints (Higginson, et al., 2007). It is possible that the constraints alter the trunk motion of the front transferor in such a way that he or she is unable to support the transferee to the same extent as during an unconstrained transfer, subjecting the rear transferor to greater loading. This hypothesis is supported by research that shows a decrease in the amount of weight able to be lifted when lifting tasks involve trunk rotation (Cheng and Lee, 2003). Unfortunately, this relationship between the presence of constraints during dependent transfers and the perception of exertion among front and rear transferors cannot be corroborated as there are currently no studies that have looked at the effect of constraints on two-person patient handling tasks.

The potential transfer of load from the front to the rear transferor during transfers in a constrained environment has implications on injury risk to the person being transferred. The true effect of the surrounding constraints on the peak vertical accelerations experienced by the transferee may have been masked given the relatively light weight of the transferees used in the current study, particularly if a shift in load between the front and rear transferor does in fact exist. As long as the rear transferor has the strength capability to lift this additional weight in a controlled manner, the difference in experienced acceleration between constraint conditions should be minimal, as shown in the current study. However, it is possible that during transfers of individuals exceeding the 50th percentile male used in the current study, the amount of weight transferred to the rear transferor in the constrained condition may exceed the strength capability to perform the transfer with the same level of control. This compromised ability to control the additional weight could potentially lead to greater accelerations experienced by the transferee in the constrained condition than were observed in the current study using a 50th percentile male.

Although transfers performed using the larger transferee, or in the presence of constraints, were perceived to be more strenuous by the front and rear transferors, vertical acceleration experienced by the transferee was not shown to increase for either of these conditions. There was also found to be no association, across all conditions, between the peak acceleration experienced by the transferee and the perceived exertion of the transferors. These results suggest that, with increased difficulty of the transfer task, both transferors chose to exert themselves to a greater extent during the task rather than perform the transfer in a less controlled manner. The opposite was found to be true regarding the direction of transfer. The perceived exertion did not differ between inboard and outboard transfers for either transferor, however greater acceleration was experienced by the transferee for the inboard direction. This lends further support to the idea that the greater potential injury risk during inboard transfers is more a function of equipment than task difficulty or mishandling by the transferors.

One of the primary limitations of this study is that, due to the novelty of the current approach used to investigate the risk factor related to injury of the transferee during two-person dependent transfers, these objective measures are not directly comparable to the subjective measures currently reported in the literature. Although the use of an anthropometric dummy facilitated the attachment of an accelerometer to record the vertical acceleration experienced by the transferee, subjective measures of comfort and security were not able to be analyzed and compared to others. Future work needs to be conducted to determine what values of peak vertical acceleration put an individual at risk. As with lifting tasks, loading frequency may play a role in the risk of injury and should also be investigated. Peak vertical accelerations observed in the current study may prove acceptable when applied infrequently, however it is possible that, due to the increased number of transfers a traveler with disabilities is subjected to during air travel, even these lower accelerations may result in increased risk of injury if the loading frequency is increased.

The current results suggest that dependent transfers of larger individuals within the spatial constraints of an aircraft are more difficult for the transferors, however the potential for injury to the person being transferred remains relatively unaffected. Also, due to differences in seating surfaces, there exists a need to perform inboard transfers in a more careful manner. Finally, the large variability observed in the acceleration experienced by the transferee indicates that some groups are less careful than others while performing dependent transfers.

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Table 4.1. Influence of dummy size on peak vertical acceleration and perceived exertion of the front and rear transferors (Mean ± SD).

	Size	
	Small	Large
Acceleration (g)	0.44 ± 0.19	0.40 ± 0.25
RPE – Front *	10.2 ± 1.8	12.2 ± 2.0
RPE – Rear *	10.6 ± 2.0	12.6 ± 2.0

* $p < 0.001$

Table 4.2. Influence of constraint on peak vertical acceleration and perceived exertion of the front and rear transferors (Mean ± SD).

	Constraint	
	Constrained	Unconstrained
Acceleration (g)	0.42 ± 0.23	0.41 ± 0.18
RPE – Front †	12.3 ± 2.1	10.1 ± 1.9
RPE – Rear *	12.1 ± 2.2	11.1 ± 2.0

* $p < 0.005$; † $p < 0.001$

Table 4.3. Influence of direction of transfer on peak vertical acceleration and perceived exertion of the front and rear transferors (Mean ± SD).

	Direction	
	Inboard	Outboard
Acceleration (g) *	0.49 ± 0.25	0.35 ± 0.19
RPE - Front	11.1 ± 1.9	11.3 ± 1.8
RPE - Rear	11.5 ± 2.0	11.7 ± 2.0

* $p < 0.005$

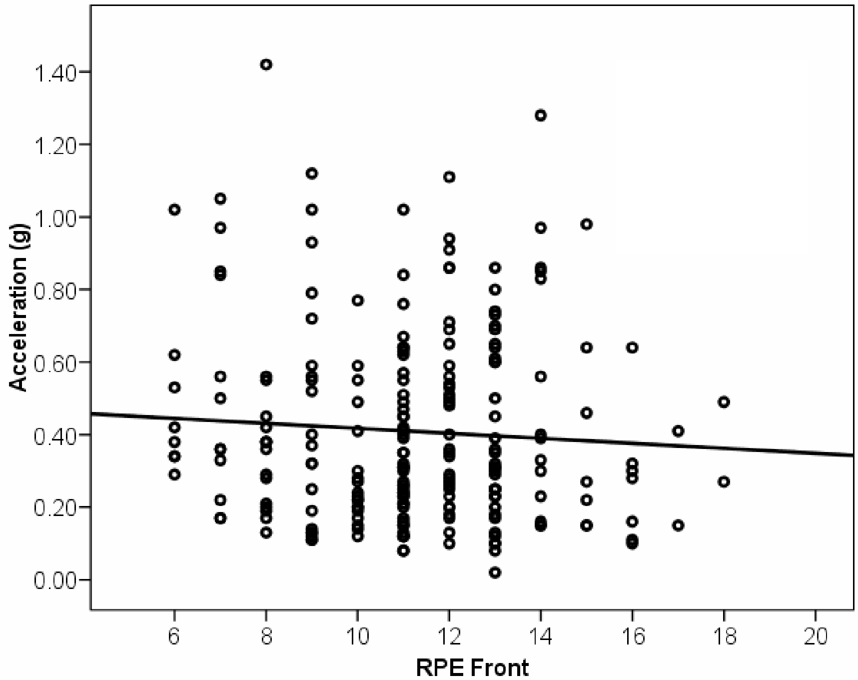

Figure 4.1. Relationship between the peak vertical acceleration of the transferee and the perceived exertion (RPE) of the front transferor. (r^2 = 0.004; p = 0.331)

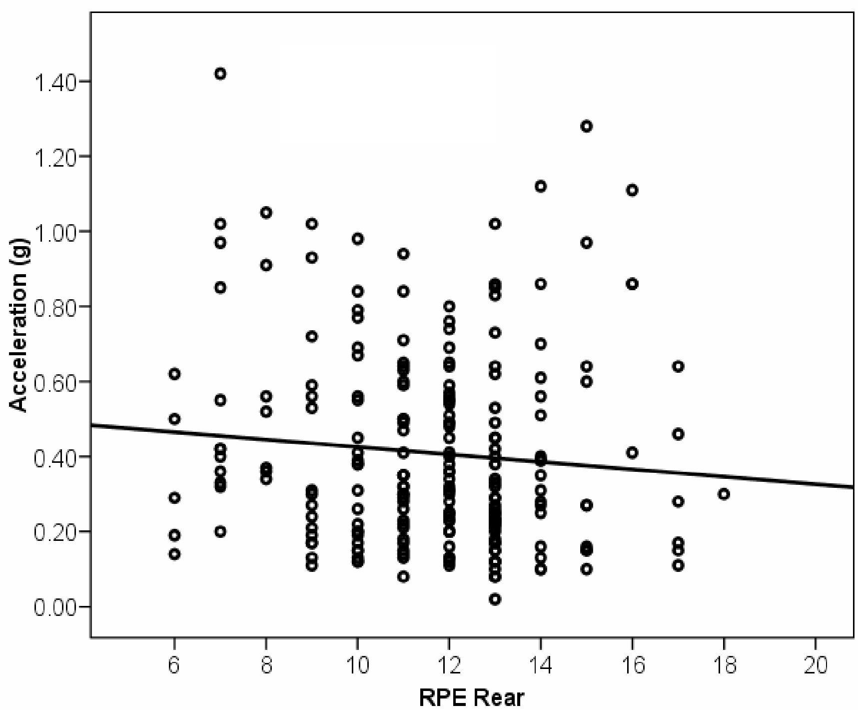

Figure 4.2. Relationship between the peak vertical acceleration of the transferee and the perceived exertion (RPE) of the rear transferor. (r^2 = 0.008; p = 0.176)

5. CONCLUSIONS

Patient transfers in clinical and assisted care settings have been associated with an increased risk of low back disorders. Although somewhat similar in nature, transfers of travelers with disabilities on board aircraft had yet to be investigated. Due to the spatial constraints imposed by the interior of the aircraft, it was reasonable to expect that the transferors performing these transfers were subjected to an increased risk of injury. The goal of the current study was to determine the effects of the spatial constraints imposed by the interior of an aircraft, transferee size, and the direction of transfer on trunk motion factors and lumbar loading (compression forces and moments) during two-person dependent transfers of a traveler with disabilities on board an aircraft. Also of interest was the perceived task difficulty of the transferors, and what influence this may have on the safety of the traveler being transferred.

As seen in other patient handling tasks, the flexion angles of both front and rear transferors while performing these transfers were found to be well above those associated with jobs known to result in high risk of low back disorders. Peak flexion of the rear transferor increased while transferring the small transferee, likely due to its shorter stature. The two-person transfer task performed in the current study appears to be primarily a sagittally symmetric lifting task, involving little out of plane trunk motion (i.e. lateral bending and twisting). The greatest number of increases in known trunk motion risk factors was observed in the front transferor while performing transfers with the large transferee, and during transfers in which the constraints were present. Although statistical analysis revealed an increase in some of these measures as a function of the imposed interventions, given the relatively small increases in these measures most would not be considered practically significant if analyzed individually. However, given what is known about complex spinal loading, these small increases taken in combination may lead to an increased risk of injury, particularly while supporting a heavy load.

Although the largest increase in moment arm length was observed in the front transferor during transfers performed in the presence of constraints, the rear transferor consistently had the largest moment arm of either transferor for all trials. This large moment arm, combined with a high trunk flexion angle and a larger percentage of supported load, would tend to put the rear transferor at a much greater risk of low-back injury while performing these transfers. Larger moment arms were associated with taller transferors, regardless of transfer position. Transferor height was also found to have a significant effect on trunk motion factors for the rear transferor, with shorter transferors demonstrating greater trunk twisting, as well as lateral bending and twisting velocities. The exact relationship of these risk factors on injury risk given transferor height remain unclear, but would appear to put taller transferors at greater risk of injury.

The constraints imposed by the interior of the aircraft and the direction of transfer had relatively little influence on lumbar loading. A small increase in resultant compressive force was observed during outboard transfers, possibly a result of a modification in lifting posture due to the high aircraft seatback. The size of the individual being transferred had the most notable influence on lumbar loading, increasing resultant compressive force, and extension, bending, and twisting moments. These results indicate that transfers of larger individuals put the rear transferor at greater risk of low back injury during a dependent transfer on an aircraft, whereas surrounding spatial constraints and the direction of transfer have little effect. The potential for injury also appears to be greater during transfers performed by smaller transferors.

Dependent transfers of larger individuals, and within the spatial constraints of an aircraft, were perceived to be more difficult by the transferors. However, the acceleration experienced by the transferee during the transfer remained relatively unaffected. This finding indicates that the transferors elected to perform the transfer in a safer manner for the transferee, regardless of the increase in task difficulty. The variability associated with the acceleration of the transferee suggests that some groups used greater care during the transfers than others. Inboard transfers resulted in greater accelerations experienced by the transferee, possibly due to differences in seating surfaces.

These results indicate that the greatest contributors to an increased risk of low back injury during dependent transfers on an aircraft include the spatial constraints of the aircraft interior, the size of the individual being transferred, and the individual characteristics of the persons performing the transfer. Obviously these factors need to be considered when developing interventions aimed at reducing these risks.

Modifications in transfer technique, transferor training, and aircraft design may contribute to a safer working environment. The high seatback of the aircraft seat appears to play a role in the risk of low back injury. Modifications to the seat height may allow taller transferors, who are able to reach over the seatback during the transfer, to maintain a more upright posture. This would also facilitate a more sagittally symmetric lift for shorter transferors who would normally be required to bend and twist around the side of the seat in order to perform the transfer. Similar interventions warrant investigation to reduce the out of plane trunk motions of the front transferor attributed to the seat located in front of the traveler.

With the possible exception of a mechanical lift located on the aircraft, interventions aimed at reducing the effect of transferee size on the lumbar loading of the rear transferor are limited. Modifications in seatback height may also serve to decrease this loading by facilitating a more upright lifting posture and decreased moment arm.

The method used to calculate lumbar compression in the current study likely underestimated the true compressive forces experienced by the lumbar spine given the high flexion angles associated with dependent transfers. The current model does not take into account the muscle activation required to provide the extension torque to maintain a flexed position while supporting a heavy external load. This muscle activation would result in elevated spinal compression values. Studies using a more complicated model, taking into account muscle activation, need to be conducted to better understand the risk of injury given the loads and flexion angles observed during a dependent transfer on an aircraft.

Given the significance of load moment on the risk of low back injury, the ability to quantify the load distribution between the front and rear transferor is important. The current study was limited to describing the changes in load moment arm of each transferor. In order to more accurately asses the potential risk of injury to the front transferor during dependent transfers, the lumbar loading experienced by the front transferor is required. Lumbar force and moment data were measured on the rear transferor. However, without information regarding the contribution of load carriage by each transferor (force applied at the hands), load moments in the current study had to be estimated. Knowledge of the load distribution between the front and rear transferor would have enabled a more accurate analysis of kinetic risk factors during dependent transfers.

It is unclear whether the variability observed in the peak vertical accelerations experienced by the transferee between groups is a result of the inexperience of the transferors in performing the required transfer task, or perhaps a result of the use of an anthropometric dummy instead of an actual person. Future research is needed to determine if similar variability is observed during transfers performed by experienced transfers, which would indicate a need to modify training protocols provided to the transferors. It is also possible that this observation is simply a function of carelessness on the part of the transferors knowing the individual being transferred was not an actual person.

The higher vertical accelerations experienced by the transferee may also be a result of transfers performed by groups with less strength capability, necessitating the inclusion of individual characteristic considerations when developing intervention strategies. The variability in transfer techniques used by the transferors in the current study may also be a function of task novelty, or may simply be adaptations made by the transferors in order to compensate for height or strength deficiencies required to perform the task safely given the heavy load and spatial limitations.

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APPENDICES

APPENDIX A

LITERATURE REVIEW

LITERATURE REVIEW

Injuries to the back are one of the most common and costly types of work-related injuries, accounting for nearly 20% of all injuries in the workplace, and nearly 25% of the annual workers compensation payments (National Safety Council, 1990). The cost associated with these low back injuries has been reported to be in excess of $20 billion annually (Pope et al., 1991), with the cost of disorders involving low back pain (LBP) estimated to be upwards of $72 billion annually (DeLuca, 1997). Of the people suffering from low back pain, over 60% attribute the condition to overexertion. Overexertion injuries in the U.S. accounted for about ¼ of all reported occupational injuries, with some industries reporting over ½ of total injuries being due to overexertion (Waters et al., 1993). A major source of work-related LBP and impairment include occupations involving manual materials handling and lifting tasks (Waters et al., 1993). Obviously there exists a need to prevent or reduce occurrence of lifting-related low back pain and injury.

This literature review will provide an overview of biomechanical risk factors associated with increased risk of low back injury during lifting tasks, as well as provide background and rationale for the current lifting guidelines established by the National Institute for Occupational Safety and Health (NIOSH). Current biomechanical models used to predict the risk of low back injury given the biomechanical risk factors will also be discussed. Finally, because statistics on the incidence of low back injury specifically relating to transfers on board aircraft are not readily available, this review will provide a brief overview of work done to date investigating patient-handling tasks in clinical and assisted care settings.

Biomechanical Risk Factors

Several biomechanical factors have been identified that may precipitate low-back pain or injury. Kinematic variables associated with increased risk of injury include lumbar flexion, lateral bending, and twisting angles, and lumbar flexion, lateral bending, and twisting velocities. (Marras et al., 1993; Norman et al., 1998; Punnett et al., 1991; Fathallah, 1998). Kinetic variables include forces crossing the lower lumbar spine (compressive, anterior-posterior shear, and medial-lateral shear forces) and lumbar moments about the three principle axes of motion (flexion, lateral bending, and twisting moments) (Herrin et al., 1986; Marras et al., 1993). Although an increase in any single kinematic or kinetic variable leads to a subsequent increase in low back injury risk, a concomitant increase in a combination of multiple factors has been found to be most predictive of injury risk (Granata & Marras, 1999; Marras et al., 1993). In addition to these biomechanical variables, greater external loads in the hands (Norman et al., 1998; Waters et al., 1993; Snook & Ciriello, 1991) and longer load moment arms (horizontal distance of the load from the spinal segment during a lift) have been found to be perhaps the most sensitive contributors to an increased risk of injury (Marras et al., 1993; Anderson et al., 1976).

Initial attempts to assess risk of injury during occupational tasks used the average values of these biomechanical variables. More recent work has shown that peak values may be more appropriate for use in discriminating injury risk (Norman et al., 1998; Herrin et al., 1986; Van Dieen & Toussaint, 1997). Epidemiological studies using peak kinematic variables such as torso angles beyond 20 deg (Punnett et al., 1991) or torso velocities (Marras et al., 1993) have been able to discriminate between high risk and low risk jobs. Peak kinetic variables including spinal compression forces and lumbar moment have also been used to classify jobs as high or low risk for low-back injury. Furthermore, damage to spinal structures as a result of excessive peak compression forces has been reported from in vitro studies (Jager et al., 1991; Adams & Hutton, 1983).

Previous research relating spinal loading to injury has focused primarily on compressive loading of the spinal segments during sagittally symmetric lifting tasks, as this was believed to result in the greatest risk of injury (Granata & Marras, 1999; Schultz & Andersson, 1981; Freivalds et al., 1984). More recent research has found that compressive loading of the spine may be greater when lifting tasks include out-of-plane movements such as twisting and lateral bending. Finite element analysis (Shirazi-Adl, 1986), as well as in vitro studies (Gunzberg et al., 1991; Shultz et al., 1979; Berkson et al., 1979), also indicate that spinal segment failure occurs at lower compressive loads when combined with torsional or shear loads. In addition, asymmetrical lifting has been found to increase lateral flexing and twisting torques (Kingma et al., 1998) and increase co-activation of antagonistic trunk muscles (Marras & Mirka, 1992). Ignoring co-contraction is undesirable in that the estimates of spine compression could be underpredicted by 45% and estimates of spine shear loading could be under-estimated by as much as 70% (Marras & Granata, 1997). These findings are supported by many epidemiological studies that have identified axial twisting and asymmetrical trunk motions involving shear and torsional spinal loading as risk factors for occupationally related low back disorders (Andersson, 1981; Kelsey et al., 1984; Kyserling et al., 1988; Punnet et al., 1991; Marras et al., 1993; Fathallah et al., 1998). Similarly the U.S. Department of Labor (1982) reports that twisting and turning is associated with a LBD event in 33% of workers. There is evidence to suggest that the injurious effects of these complex dynamic motions may be dependent upon the degree of sagittal flexion present when these bending and twisting motions occur. While looking at postural and dynamic motion patterns of industrial workers to determine LBD risk, Fathallah et al. (1998) found that combined lateral and twisting velocities were similar across all three risk groups when sagittal flexion angles were less than 15 degrees. However, when flexion angles exceeded 15 degrees, combined lateral and twisting velocities were able to distinguish between risk groups.

Established Lifting Guidelines

In 1981, NIOSH established lifting guidelines in an effort to prevent or reduce the occurrence of lifting-related low back pain. These guidelines resulted in a lower level 'Action Limit' of 3400 N of spinal compression, defined as being protective of 90% of workers (99% of males and 75% of females) for occasional lifting tasks (less than three lifts/min). An upper 'Maximal Permissible Limit' of 6400 N was established and defined as being 'hazardous to most workers' (NIOSH, 1981). This upper limit is representative of loads that 25% of males and only 1% of females can safely lift. These guidelines were based on results from cadaver as well as epidemiological studies.

In the cadaver studies, isolated spines were subjected to a combination of compression and moment loading conditions until the specimens failed (Evans and Lissner, 1965; Sonoda, 1962; Perey, 1957). A summary of results from 25 such cadaveric studies (Jager, 1987) shows great variability, with average compression failure values ranging from well below the Action Limit (around 2000 N) to well above the Maximal Permissible Limit (approximately 8000 N). These results were similar to those of Brinckman et al. (1988), who found compressive strength values to range from 2100 to 9600 N, with 21% of cadaver spinal segments experiencing end-plate failure or fracture at loads below 3400 N. Jager and Luttman (1989) found similar failure rates in 307 lumbar segments, with 30% of spinal segments failing below 3400 N.

Epidemiological studies have been used to relate the incidence of LBP in industrial jobs, such as manual materials handling tasks, to estimates of vertebral compressive forces predicted from biomechanical models. Incidence of LBP has been found to be 1.5 times greater for jobs with predicted compressive force between 4500 and 6800 N than for jobs with compressive forces lower than 4500 N (Herrin et al., 1986). In another study, muscular strains were found to occur in workers experiencing average estimated compressive forces of 5340 N, whereas disc injuries were more likely to occur at compressive forces greater than 8000 N (Bringham and Garg, 1983). Similarly, Anderson (1983) found that the incidence of LBP increased by 40% in male workers performing lifting tasks in which predicted compressive loads exceeded 3400 N.

The primary limitation of the 1981 lifting equation was that it could only be accurately applied to lifting tasks performed in the sagittal plane. For this reason, the original lifting equation was revised in 1991 to provide guidelines for a greater range of lifting tasks, including asymmetrical lifting tasks, and tasks involving the lifting of objects with less than optimal couplings (hand-object interfaces, i.e. handles). The revised equation resulted in the development of the Recommended Weight Limit, defined as the weight of the load that nearly all healthy workers could lift regularly over a substantial period of time (8 hours) without an increased risk of developing a lifting-related low back disorder (NIOSH, 1994). This Recommended Weight Limit is a function of the horizontal location of the object being lifted, vertical location of object being lifted, vertical travel distance of object being lifted, angle of asymmetry, lifting frequency, and coupling class (quality of hold on object). Although superior to the 1981 equation, the revised equation still possesses limitations. It is not meant to be used for tasks which involve lifting or lowering in a restricted workspace, or lifting or lowering of unstable objects, such as those unable to maintain shape or balance point throughout the duration of the lift.

The NIOSH lifting equations and tolerance limits are currently the most convenient means to quickly assess injury risk severity for a given lifting task and, as such, are used extensively in the literature to quantify the risk associated with a broad range of lifting tasks. In particular, the NIOSH lifting equations provide a convenient tool for estimating safe lifting loads in the workplace and for guiding changes in workplace design. However, for lifting tasks in which the load and relevant workplace variables are fixed, such as the dependent transfers used in the current study, it is often desirable to instead compare the spinal loading experienced by the transferors to known tolerance limits. In order for tolerance limit comparisons to be made, an estimate of spinal loading must be established. This is most commonly achieved through the use of biomechanical models.

Biomechanical Models

In order to understand the mechanisms responsible for occupationally-related LBD, we must be able to accurately assess the loading that occurs at the most common site of injury, the lower lumbar spine, during whole-body, free, dynamic lifting conditions. Through studies of loads, distances, muscle activities, pressures, and external forces, it is possible to calculate stresses acting on different spinal elements. An accurate evaluation of spinal loading is necessary in order to compare loads imposed upon the lumbar spine to tolerance limits derived from cadaver studies, epidemiological studies, and finite element models of the spine. The evaluation of loads on the lumbar spine is a rather difficult task. Unfortunately, at the present time, there is no feasible method whereby such data can be measured directly. Direct measurement techniques, such as intra-discal force transducers, are expensive and dangerous due to their invasive nature, particularly when motion is involved.

For this reason, several biomechanical analysis techniques (models) have been developed that are capable of predicting loads on the lumbar spine based on Newtonian laws of physics. Biomechanical modeling has become an increasingly popular tool for estimating loads on the lumbar spine and can be accomplished using many different techniques. The most common strategies used to estimate external loads on the lumbar spine include simple static or quasi-static techniques, or more advanced dynamic techniques. These estimates of external loading are calculated from information about body position and the external forces acting on the body and provide an approximation to the net effect of all internal forces acting across a given joint. Estimation of internal loads on individual tissues requires the use of more complex techniques, including EMG-assisted models, computer optimization techniques, and regression equations that predict loads based on such variables as kinetic or kinematic body motion information, workplace factors, and load characteristics.

Due to their simplicity and ease-of-use, most current lumbar spine models are static (i.e. effects of motion are neglected) and limited to analysis of motions occurring only in the sagittal plane (Chaffin et al., 1977; Ekholm et al, 1982; Roozabar, 1975; Schultz et al., 1982). Although adequate for the analysis of relatively slow lifting tasks (Garg et al., 1982), it has been well documented that these models consistently underestimate spinal loading during dynamic lifting tasks by as much as 20-60% (Marras et al., 1998; Leskinen et al., 1983; Marras & Sommerich, 1991). McGill and Norman (1985) reported peak moments to be 19% higher on average, with a maximum difference of 52%, using a dynamic model compared to a static model. More drastic differences were found by Garg et al. (1982), who found compressive forces and peak moments to be two to three times greater using a dynamic versus static approach.

Quasi-static models have been used when the inertial effects due to the acceleration of the load greatly exceed those due to the motion of the body segments (McGill & Norman, 1985; Gagnon et al., 1986). Incorporating the effects of the acceleration of the load into an otherwise static model, the load moments were found to be 25% greater than those predicted using a fully dynamic model (McGill & Norman, 1985). Skotte et al. (2002) reported a similar difference of 17% between dynamic and quasi-static estimations in an isolated case, but found less than 5% difference in most cases (90%).

Although more computationally intensive than static and quasi-static models, dynamic models that incorporate all inertial effects give the most accurate estimate of spinal loading. However, they are often limited to analysis in a laboratory setting due to the equipment required to accurately record the force and position data needed to solve the equations of motion. Another limitation associated with these models is that the spinal loads predicted from the external loading typically fail to fully account for the contributions of anatomical structures and musculature. In order to increase spinal stability during motion, co-contraction of antagonistic lumbar musculature has been found to increase during lifting tasks, particularly during dynamic, out-of-plane movements (Granata & Marras, 1995). Failure to account for this additional loading associated with co-contraction can result in the underprediction of compressive forces by as much as 45%, and of shear forces by as much as 70 % (Marras & Granata, 1997).

Regression modeling has become an increasingly popular tool to estimate loads placed on the lumbar spine during lifting movements from combinations of variables such as workplace factors, load characteristics, and biomechanical factors (kinetic and kinematic variables). Since these equations are based on easily measured variables, they have the potential to be extremely useful to ergonomists in evaluating injury risk in industrial and field environments.

One of the first attempts to predict spinal compression using easily acquired variables was that of Susnik and Gazvoda (1986). Using a static, sagittally-symmetric model, they were able to adequately predict spinal compression from object weight, trunk angle, and upper body weight during “back lifting” tasks. Prediction equations were developed for cases in which there existed both small ($R^2 = 0.92$) and large ($R^2 = 0.95$) load moment arms. Obviously this model is limited in its usefulness for ergonomic applications, as it is not representative of most lifting tasks and it is based on a static analysis, which has been shown to underestimate compressive forces during dynamic activities.

Potvin et al. (1992) improved on these initial prediction equations by incorporating the inertial components associated with lifting tasks through the use of a dynamic analysis. Their results indicate a strong correlation ($R^2 = 0.93$) between predicted peak compressive force from the dynamic analysis and four variables measured at the time of peak hand force: subject mass, hand force, moment arm from the load to the ankle, and moment arm from the load to L4/L5. A ‘field’ model was also developed that utilized variables that could be easily obtained in a working environment without relying on the use of specialized equipment. The result was a seven-variable model that was able to predict peak compression forces using load mass, subject weight, subject abdominal girth, pelvic and trunk angles, lift technique (stoop or squat), and the horizontal distance from the load to the ankle ($R^2 = 0.90$). Although an improvement over the Susnik and Gazvoda (1986) model, it is still limited to assessment of lifting tasks performed in the sagittal plane.

More recent work has utilized workplace characteristics, as well as kinematic and kinetic variables, to predict injury risk during lifting tasks in the workplace (Marras et al., 1993; McGill et al., 1996). McGill et al. (1996) developed a four-dimensional regression equation using low back compression and the three-dimensional moments predicted about L5/S1 from a 90-muscle EMG-assisted model (R^2 = 0.937). Unlike previous models, the McGill model takes into account muscle coupling and co-contraction, and is able to predict compressive forces during out-of-plane movements, a task commonly found in many industrial and manual materials handling environments.

Marras et al. (1993) used an epidemiological approach to determine which workplace and biomechanical factors were best able to discriminate between jobs that placed workers at high or low risk of LBD. Multiple logistic regression was used to predict the probability of group membership (high or low risk) as a function of various workplace and trunk motion factors. The five factors best able to discriminate between high and low risk group membership (odds ratio: 10.7) included lifting frequency, maximum load moment, maximum trunk lateral velocity, average trunk twisting velocity, and maximum trunk flexion angle. This analysis also provides additional evidence that the risk of low back injury is not a function of any one single factor, but many factors in combination. With the exception of maximum load moment, average sagittal velocity resulted in the greatest odds ratio (3.33), indicating that this variable is the single best trunk motion variable for discriminating between risk groups. Interestingly this variable was not included in the final risk model. Of the variables included in the final model, none were able to reliably discriminate between high and low risk situations. However, when considered in combination, the mean differences in these five variables between groups at low and high risk were associated with a nearly eleven-fold difference in the odds of high-risk group membership.

With the use of kinetic and kinematic variables as predictors of outcome measures such as lumbar compression force, it is of interest to determine if one method is inherently more accurate at predicting such outcomes than another. In analyzing symmetric and asymmetric dynamic lifting tasks, Fathallah et al. (1999) developed three classes of regression models to predict spinal loading (compression, lateral shear, and anterior-posterior shear) as estimated from an EMG-assisted model. The three model classes consisted of kinematic, kinetic, and combined models. The kinematic model utilized subject (body weight), workspace (box weight), and trunk motion (sagittal, lateral, and twisting position, velocity, and acceleration) parameters. The kinetic model consisted of the same subject and workspace parameters, with the addition of the three measured lumbar moments. All variables were used in the combined model. The results indicated that peak spinal loading could be predicted equally well using any of the three models. Continuous loading appeared to be most accurately estimated using the combined model, although the kinematic model was adequate. Model selection also appeared to be dependent on lift symmetry and the weight of the object being lifted. The predictive ability of all three models tended to improve for the compression measure as the weight being lifted increased. During asymmetric lifting, the kinematic model performed better under low-weight conditions, whereas the kinetic and combined models seemed to perform better as the weight increased.

Although most of these models possess inherent limitations, they are the best tools available for estimating spinal loads during tasks such as the transfers performed in the current study. In this study, we plan to use the models of Marras et al. (1993) and McGill et al. (1996) to estimate the loads experienced at the level of the lumbar spine, and compare these loads to known tolerance limits. With this information, it will be possible to detect changes in injury risk given various lifting conditions.

Patient Handling Literature

As mentioned previously, there exists a paucity of information regarding the prevalence of low back injuries during dependent transfers of disabled travelers on board aircraft, as well as quantitative research addressing the potential for injury while performing such a task. Due to the high prevalence of occupational back injuries associated with patient-handling tasks in clinical and assisted care settings, research in this area is more prolific. Most patient handling tasks can be broken down into three categories: lifting/transferring, repositioning, and turning (Skotte et al., 2002). Therefore, the current literature typically focuses on comparison of injury risk among the different patient-handling tasks (Schibye et al., 2003; Daynard et al., 2001; Skotte et al., 2001, 2002), influence of different techniques on the same task (Varcin-Coad & Barrett, 1998; Winkelmolen et al., 1994), comparison of one- and two-person transfers given the same task (Marras et al., 1999a; Garg et al., 1991), or the difference between manual transfers and the use of mechanically-assisted transfer devices for the same task (Ulin et al., 1997; Garg et al., 1991).

Despite the wide scope of the literature, information regarding potential injury risk during two-person dependent transfers is limited to very few patient-handling studies (Daynard et al., 2001; Marras et al., 1999a; Ulin et al., 1997; Garg et al., 1991) and even fewer two-person manual materials handling studies (Karwowski, 1998; Karwowski & Mital, 1986; Karwowski & Pongpatanasuegsa, 1988; Marras et al., 1999a). Of the relevant literature, there seems to be a general consensus that such a patient transfer task has the potential to result in a high risk of injury.

Most two-person patient transfers result in estimates of compression forces that fall between the 3400 N Action Limit and the 6400 N Maximum Permissible Limit established by NIOSH. Garg and Owen (1991) found compression forces of 4751 N in nurses during two-person patient transfers. Similar forces (4400 N) were found by Gagnon et al. (1986) and Garg et al. (1991) (4464-5414 N) using a two-dimensional static model and by Skotte et al. (2002) (3313-5162 N) using a three-dimensional dynamic model. Using a dynamic model, Marras et al. (1999b) found slightly higher compression values when looking at two-person transfers between hospital beds, wheelchairs, hospital chairs, and commode chairs. Peak spinal compression in approximately 15-20% of these two-person transfers fell above the 6400 N tolerance limit. For all two-person transfers investigated, transfers between the hospital chair and commode chair had the highest LBD risk (87%) and compression (5178 N) associated with them. Ulin et al. (1997) found the highest compression forces, ranging from 6000-6500 N, in female nurses while transferring patients between a bed and wheelchair. The magnitude of these forces depended on the size of patient, and the height and position of the transferor. Peak compression forces were computed to be as high as 9292-12936 N for tall lead nurses transferring a heavy patient (96 kg). Although the details of the transfer technique used by Ulin et al. (1997) are rather vague, they appear to be the only study to use a two-person "front and rear" technique, similar to the one used in the current study. All other studies using a two-person transfer technique have been performed with both transferors positioned in front of the person being transferred.

High shear loads have also been associated with patient transfer tasks. Using a static biomechanical analysis Garg et al. (1991) estimated shear loads of 934 N while transferring patients from a shower chair to a wheelchair using a two-person "in front" transfer technique. Using the same transfer technique, Marras et al. (1999b) found lateral shear forces to range from 660-817 N, and anterior-posterior shear forces ranging from 913-950 N during transfers between hospital beds, wheelchairs, hospital chairs, and commode chairs. While these shear forces are just below the 1000 N limit found to cause damage to the annulus fibrosus by Farfan (1988), the Marras et al. (1999b) study showed that 19-42% of all two-person transfers exceeded this shear load tolerance limit.

These studies indicate that patient handling tasks have the potential to result in serious injury to the persons performing the transfer, with spinal loads approaching, or sometimes exceeding, known tolerance limits. Although the patient handling literature provides insight into the spinal loadings we may expect to see in the current study, our study hopes to expand on this information by including components not previously investigated. With the possible exception of Ulin et al. (1997), the front and rear transfer technique used in the current study has not previously been investigated, and their study was of transfers to and from a bed, not a seat. Similarly, the effects of constraints while transferring a patient have yet to be studied. Few studies have looked at the effects of transferee weight on injury risk to the transferors, and none of these required the transfers to be performed around or over constraints.

APPENDIX B

INFORMED CONSENT DOCUMENT

INFORMED CONSENT DOCUMENT

Project Title: Biomechanics of Dependent Transfers on an Aircraft
Principal Investigator: Michael Pavol, Ph.D., Dept. of Exercise & Sport Science
Research Staff: Brian Higginson, Chris Schafer, Brad Black

PURPOSE

This is a research study. The purpose of this research study is to investigate how different factors associated with the transfer of a traveler with disabilities between a wheelchair and an aircraft seat influence the potential for injury. Such "dependent" transfers of those who cannot do so themselves are the major source of injury to both traveler and airline staff during commercial air travel by people with disabilities. The traveler is at risk of being dropped or of a shoulder injury, while the transferors are at risk of low back or other musculoskeletal injury. There is a need to reduce these risks. This study will therefore investigate how the design of an aircraft interior interacts with the individuals involved in a dependent transfer to influence the risk of injury to the transferors and to the traveler. We expect that this research will lead to specific design and procedural modifications to increase transfer safety. The purpose of this consent form is to give you the information you will need to help you decide whether to be in the study or not. Please read the form carefully. You may ask any questions about the research, what you will be asked to do, the possible risks and benefits, your rights as a volunteer, and anything else about the research or this form that is not clear. When all of your questions have been answered, you can decide if you want to be in this study or not. This process is called "informed consent". You will be given a copy of this form for your records.

We are inviting you to participate in this research study because you are a healthy adult between 18 and 40 years of age, you are experienced in and accustomed to lifting over the past month through regular strength training or the equivalent, and you have no history of low back pain. Approximately 100 subjects will participate in this study.

PROCEDURES

If you agree to participate, your involvement will last for one session of approximately 1.5 hours. The following procedures are involved in this study.

- **Health History Questionnaire:** You will record your health history on a questionnaire that will take approximately 5-10 minutes to complete. It is possible that we may ask you not to participate in this study as a result of the information you provide on this questionnaire.

- **Instruction in Transfer and Lifting Techniques:** We will teach you how to perform the standard two-person transfer technique that is used on an aircraft. One transferor works from behind, reaching through the traveler's armpits and grabbing hold of his or her forearms. The second transferor works from in front, grabbing the traveler beneath the thighs at the knees. On a verbal signal, both transferors lift up together and shift the traveler sideways from the wheelchair to the aircraft seat or vice versa. The proper lifting techniques for reducing the risk of low back injury will also be described and demonstrated. If you have any concerns about your participation in the study, at that or any other time, you must tell us immediately.

- **Warm-up Exercises:** You will perform a set of warm-up exercises to prepare your muscles for lifting. The warm-up will begin with 5 minutes of brisk walking. This will be followed by a set of squats and a set of stiff-legged deadlifts with just your body weight for resistance. You will repeat these exercises while holding a 10 lb. weight in front of you, then while holding a 25 lb. weight in front of you.

- **Practice in Transferring:** For the rest of the testing, you will work with another subject as your partner to transfer a dummy between a standard aircraft aisle wheelchair and a standard aircraft seat. You and your partner will decide between yourselves which transferor position each of you will take, front or back, during all of the transfers that you will be performing. In deciding, be aware that the transferor in back will have to lift an estimated 65-75% of the weight of the dummy, while the transferor in front will have to lift only 25-35% of the weight. It is therefore likely that the transferor in back will be at greater risk of injury. You must be willing to participate in either transferor position. The transferor in back will put on a heart rate monitor. You and your partner will then perform two practice transfers using the standard two-person technique described above. One of two dummies will be used: medium or small. The medium dummy weighs 172 lb., of which we estimate the transferor in back will need to lift 112-129 lb. and the transferor in front 43-60 lb. The small dummy weighs 125 lb., of which we estimate the transferor in back will lift 81-94 lb. and the transferor in front 31-44 lb.

- **Force and Motion Measures During Transfers**: You will wear athletic shoes, shorts, and a tank top or T-shirt. Next, about 34 small, reflective Styrofoam balls will be taped to your skin and clothes for our cameras to see. Loose clothing may be pinned or taped. Long hair may be placed in a ponytail. You will be filmed while standing still briefly. You and your partner will then work together to perform 8 trials in which you transfer the dummy from the wheelchair to the aircraft seat or vice versa. You will be allowed to rest for as long as you want between trials. The following factors may change between trials:
  1. *Transferee Size*: Either the small- or medium-sized dummy may be used in the transfer.
  2. *Row Spacing*: Frames made of metal pipe will be used to simulate the rows of seats in front of and behind where the transfer is taking place. We may change the spacing between rows from trial to trial, or we may remove the other rows completely for some trials.

3. *Aisle Spacing*: Frames made of metal pipe will also be used to simulate the seats across the aisle from the transfer. These frames may either be present or absent from trial to trial.
4. *Headroom*: A frame simulating the overhead bins may either be present or absent from trial to trial.
5. *Sliding Board Use*: A sliding board is a smooth board, placed between the wheelchair and aircraft seat to allow the dummy to be moved by sliding instead of lifting. This board might be used in selected trials.
6. *Seat Backrest Height*: Modifications to the aircraft seat will allow the upper half of the seat backrest to be removed for selected trials.

The conditions tested will vary from one pair of subjects to another and will be selected just before testing. During each trial, cameras will film the motion of the reflective markers taped to you and your partner. If you are the transferor in back, sensors will also measure the forces between your feet and the ground and the forces between your thigh and the aircraft seat.

- **Anthropometric Measures:** Finally, you will have your weight and selected body dimensions measured. These dimensions include your foot length, ankle width, knee width, pelvis-to-waist height, and arm width at the shoulder. Body weight will be measured using a scale. Standing body height will be measured using a type of wall-mounted ruler. All other body measurements will be made during standing using a special set of calipers designed for this purpose.

- **Follow-up:** Two to four days after your testing session, we will telephone you to find out whether you experienced any back or muscle pain as a result of participating in this study. You will be asked a series of questions about any such pain. The phone call will take less than 10 minutes.

RISKS

The possible risks associated with participating in this research project are as follows. You may experience some mild muscle soreness in the days following testing. As with all lifting activities, you could strain a muscle, particularly in your lower back, but possibly also in your arms or legs. It may be possible to rupture or herniate one of the disks in your spine. **Back injuries of this type may cause long-term disability, require treatment for months or years, and may limit lifestyle and employment for months or years.** Among other risks, you could injure a ligament in your knee or ankle if you twist it during a transfer. Vigorous exertion can also lead to chest pain or a heart attack. A number of steps have been taken to minimize the risk involved in participating in this study. First, we have limited the number of transfers that you will perform and we will allow you to rest as much as you want between transfers. We will instruct you in the proper lifting technique to minimize the risk of injury. You will also undergo a set of warm-up exercises to prepare your muscles for lifting. We will closely watch you during each transfer and will suggest how to correct any problems we see in your technique. Finally, we have tried to keep the weight being lifted from being too large. Guidelines from the National Institute for Occupational

Safety and Health (NIOSH) recommend that, to avoid injury, you should not lift more than 96 lb. in the situations being tested. The dummy you will be asked to lift will weigh up to 172 lb., but a partner will help you. It is important to know that the amount of weight that must be lifted during a transfer differs between the transferor in front and the transferor in back. We estimate that the transferor in back must lift 65-75% of the weight of the dummy, while the transferor in front must lift only 25-35% of the weight. It is therefore likely that the transferor in back is at greater risk of injury. Furthermore, for the 172 lb. dummy, the weight that must be lifted by the transferor in back is 112-129 lb., which is above the NIOSH-recommended maximum weight that should be lifted. Because of these risks, you must tell us of any concerns you have, if and when they arise. Also because of these risks, **if you experience any pain during or after a transfer, you must tell us immediately and withdraw from the study.** Finally, if you experience **any** back or muscle pain in the days following participation in this project, you **must** inform the primary investigator of this project, Michael Pavol, Ph.D., at (541) 737-5928 or by e-mail at mike.pavol@oregonstate.edu.

BENEFITS

There will be no personal benefit for participating in this study. However, the researchers anticipate that, in the future, society may benefit from this study in that it may be possible to make dependent transfers onboard an aircraft safer for travelers with disabilities and for the airline personnel who perform the transfers. If so, people with disabilities may be more willing to travel by air and thereby participate more fully in society.

COSTS AND COMPENSATION

You will not have any costs for participating in this research project. You will receive $20 as compensation for participating in this research project. If you withdraw before completing the study, your compensation will be based on the number of trials you performed: $5 for the first practice transfer, $8 for the first transfer you were filmed during, and $1 for each transfer thereafter.

CONFIDENTIALITY

Records of participation in this research project will be kept confidential to the extent permitted by law. However, federal government regulatory agencies and the Oregon State University Institutional Review Board (a committee that reviews and approves research studies involving human subjects) may inspect and copy records pertaining to this research. It is possible that these records could contain information that personally identifies you. To preserve your anonymity and confidentiality, your data will be identified only by an assigned subject code, and not by name. Only the researchers will have knowledge of your name and code number. Any documents that include your name will be stored in a locked filing cabinet in the Biomechanics Laboratory and will be accessible only to the research staff of this study. In the event of any report or publication from this study, your identity will not be disclosed.

Results will be reported in a summarized manner in such a way that you cannot be identified.

VISUAL AND MOTION CAPTURE RECORDINGS

By initialing in the space provided, you verify that you have been told that both visual and motion capture recordings will be generated during the course of this study. The motion capture recordings will be used to measure your movements during the transfers and to calculate the forces involved. Only the reflective markers that were attached to you will appear in these motion capture images; you will not be visible. The visual recordings will be used to observe your position relative to objects that markers were not attached to; you will be visible in these images. As indicated above, all recordings will be identified only by an assigned subject code. The recordings will be stored on our motion capture system or on CD's in the Biomechanics Laboratory. They will be accessible only to those working in the laboratory and will be kept no longer than 7 years. These recordings will potentially be used for other, related research projects.

_______________ Participant's initials

RESEARCH RELATED INJURY

In the event of research related injury, compensation for medical treatment is not provided by Oregon State University.

VOLUNTARY PARTICIPATION

Taking part in this research study is voluntary. You may choose not to take part at all. If you agree to participate in this study, you may stop participating at any time. You may skip any questions on the health history questionnaire that you prefer not to answer. If you decide not to take part, or if you stop participating at any time, your decision will not result in any penalty or loss of benefits to which you may otherwise be entitled. Any data collected from you will be saved and may be included in the study results, even if you choose to withdraw partway through the study. As described, if you withdraw prior to completing the study, your compensation will be pro-rated based on the number of trials performed: $5 for the first practice transfer, $8 for the first transfer you were filmed during, and $1 for each transfer thereafter.

QUESTIONS

Questions are encouraged. If you have any questions about this research project, please contact: Michael Pavol, Ph.D., at (541) 737-5928 or by e-mail at mike.pavol@oregonstate.edu. If you have questions about your rights as a participant, please contact the Oregon State University Institutional Review Board (IRB) Human Protections Administrator, at (541) 737-3437 or by e-mail at IRB@oregonstate.edu.

Your signature indicates that this research study has been explained to you, that your questions have been answered, and that you agree to take part in this study. You will receive a copy of this form.

Participant's Name (printed): ____

___________________________________ ____________________
(Signature of Participant) (Date)

There is a chance you may be contacted in the future to participate in an additional study related to this project. If you would prefer not to be contacted, please let the researcher know, at any time.

RESEARCHER STATEMENT

I have discussed the above points with the participant or, where appropriate, with the participant's legally authorized representative, using a translator when necessary. It is my opinion that the participant understands the risks, benefits, and procedures involved with participation in this research study.

__________________________________ ____________________
(Signature of Researcher) (Date)

APPENDIX C

HEALTH HISTORY QUESTIONNAIRE

Subject Code: ___________________

OREGON STATE UNIVERSITY

BIOMECHANICS LABORATORY

Health History Questionnaire

Sex: ____ Male ____ Female Height: __________________

Age: ______ Weight: __________________

Health History:

Have you ever had:	Yes	No	Explain: Current symptoms?	Yes	No
Back pain or injury					
Scoliosis (curved spine)					
Spinal cord injury					
Neck pain or injury					
Shoulder pain or injury					
Ligament injury					
Joint pain					
Muscle injury or tendonitis					
Broken bones as an adult					
Head injury					
Arthritis					
Neurological problems or conditions					

Health History (cont.):

Have you ever had:	Yes	No	Explain: Current symptoms?	Yes	No
Cancer or tumor					
Diabetes					
Heart trouble/heart attack					
Disease of the arteries					
High blood pressure					
Epilepsy					
Lung disease					
Asthma					
Surgeries/operations					
Other illness or disease					

Present Symptoms:

Have you recently had:	Yes	No	Explain:
Back pain			
Chest pain			
Cough on exertion			
Heart palpitations			
Shortness of breath			
Muscle injury or tendonitis			
Painful, stiff or swollen joints			
Pregnancy			
Other illness or disease			

Physical Activity and Exercise:

1. Have you done any strength training (e.g. training using weights or resistance machines) within the last month? *(Circle one answer, then follow arrow to next question)*

NO

YES

1A. If YES, about how many times per week, on average, have you done strength

training within the past month? *(Circle one number)*

0. Less than once per week
1. Once per week
2. Twice per week
3. More than twice per week

1B. Which of the following best describes the strength training you've done within the

past month? *(Circle one number)* (Note: max. = heaviest weight you can lift once)

1. Heavy weights (more than 85% of max.), focus on size and strength
2. Medium weights (60–85% of max.), focus on endurance and strength
3. Light weights (less than 60% of max.), focus on fitness

2. Within the last month, have you done any other physical labor that involved lifting moderate-to-heavy objects (i.e. objects weighing 25 lbs. or more) at least 10 times in one day, either at home or at work? *(Circle one answer, then follow arrow to next question)*

NO

YES

2A. If YES, about how many days per week within the last month, on average, did you

do this labor involving repeated moderate-to-heavy lifting? *(Circle one number)*

0. Less than one day per week
1. One day per week
2. Two days per week
3. More than two days per week

2B. On those days within the last month that you did this labor involving repeated

moderate-to-heavy lifting, about how many times during a day's labor did you

perform each of the following tasks, on average? *(Circle one answer in each row)*

	TIMES PER DAY		
Moderate lifting (i.e. objects weighing 25–50 lb.)	0–9	10–99	100+
Heavy lifting (i.e. objects weighing more than 50 lb.)	0–9	10–99	100+

Go to next page

3. Following is a list with examples of physical activities that are considered light, moderate, and vigorous. On average within the last month, about how many hours per week did you spend doing each type of physical activity? *(Circle one answer in each row)* (Note: do not include time spent in strength training or in physical labor involving moderate-to-heavy lifting)

	HOURS PER WEEK				
Light activity (e.g. brisk walking, housework, yoga, dancing, golf) …….……………………..	0–½	1–1½	2–3½	4–6½	7+
Moderate activity (e.g. aerobics, yard work, home repair, stair-climber, basketball) ……..	0–½	1–1½	2–3½	4–6½	7+
Vigorous activity (e.g. running, swimming, bicycling, rowing, tennis, soccer, skiing) …...	0–½	1–1½	2–3½	4–6½	7+

4. Within the past year, have you assisted in transferring, to or from a seat or wheelchair, a person who could not do so on his/her own due to disability?

   NO

   YES

Drugs and Medications:

Certain types of drugs and medications may impair your judgment, your alertness, your balance, and/or your ability to feel pain, placing you at increased risk of injury during this study. It is therefore very important that we know if you have taken any of the following drugs within the past 48 hours:

- Alcohol (2 or more beers, glasses of wine, or "hard" alcoholic drinks)
- Sedatives or anxiety/tension relief medication (e.g. Halcion, Xanax, Phenobarbital)
- Recreational drugs (e.g. marijuana, cocaine)
- Antihistamines or cold medications (e.g. Tylenol PM, Benadryl)
- Anti-inflammatory medication/pain relievers (e.g. aspirin, Ibuprofen)

Have you taken any of the types of drugs and medications listed above within the past 48 hours? *(Circle one answer)*

NO

YES

Is there any other information that you feel we should know about your health?

APPENDIX D

ANALYSIS OF VARIANCE TABLES

Kinematic Analysis

Lumbar flexion angle of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	1718.190	1	1718.190	83.56	.000
Error(size)	657.938	32	20.561		
const	136.311	1	136.311	4.416	.044
Error(const)	987.668	32	30.865		
dir	23.940	1	23.940	1.051	.313
Error(dir)	728.698	32	22.772		
size * const	53.911	1	53.911	2.775	.106
Error(size*const)	621.788	32	19.431		
size * dir	9.206	1	9.206	.585	.450
Error(size*dir)	503.602	32	15.738		
const * dir	45.086	1	45.086	2.905	.098
Error(const*dir)	496.652	32	15.520		
size * const * dir	1.686	1	1.686	.153	.699
Error(size*const*dir)	353.592	32	11.050		

Lumbar bending angle of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	53.370	1	53.370	10.455	.003
Error(size)	163.356	32	5.105		
const	7.501	1	7.501	.614	.439
Error(const)	390.880	32	12.215		
dir	72.660	1	72.660	3.864	.058
Error(dir)	601.786	32	18.806		
size * const	9.814	1	9.814	2.016	.165
Error(size*const)	155.748	32	4.867		
size * dir	.386	1	.386	.050	.825
Error(size*dir)	249.290	32	7.790		
const * dir	8.261	1	8.261	1.495	.230
Error(const*dir)	176.790	32	5.525		
size * const * dir	.536	1	.536	.145	.706
Error(size*const*dir)	118.085	32	3.690		

Lumbar twisting angle of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	.014	1	.014	.005	.946
Error(size)	93.458	32	2.921		
const	3.341	1	3.341	.619	.437
Error(const)	172.695	32	5.397		
dir	6.213	1	6.213	.674	.418
Error(dir)	295.043	32	9.220		
size * const	1.187	1	1.187	.611	.440
Error(size*const)	62.105	32	1.941		
size * dir	.418	1	.418	.160	.692
Error(size*dir)	83.444	32	2.608		
const * dir	.036	1	.036	.014	.907
Error(const*dir)	84.220	32	2.632		
size * const * dir	.017	1	.017	.006	.941
Error(size*const*dir)	96.175	32	3.005		

Lumbar flexion velocity of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	196.909	1	196.909	2.595	.117
Error(size)	2428.281	32	75.884		
const	30.818	1	30.818	.322	.574
Error(const)	3060.537	32	95.642		
dir	12390.280	1	12390.280	62.493	.000
Error(dir)	6344.495	32	198.265		
size * const	15.612	1	15.612	.249	.621
Error(size*const)	2005.583	32	62.674		
size * dir	121.367	1	121.367	.958	.335
Error(size*dir)	4053.228	32	126.663		
const * dir	10.721	1	10.721	.181	.673
Error(const*dir)	1893.059	32	59.158		
size * const * dir	44.839	1	44.839	.463	.501
Error(size*const*dir)	3097.201	32	96.788		

Lumbar extension velocity of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	463.220	1	463.220	3.349	.077
Error(size)	4426.234	32	138.320		
const	402.808	1	402.808	4.029	.053
Error(const)	3199.471	32	99.983		
dir	487.914	1	487.914	2.591	.117
Error(dir)	6025.505	32	188.297		
size * const	2.128	1	2.128	.036	.852
Error(size*const)	1913.656	32	59.802		
size * dir	.024	1	.024	.000	.988
Error(size*dir)	3220.230	32	100.632		
const * dir	10.282	1	10.282	.107	.745
Error(const*dir)	3068.247	32	95.883		
size * const * dir	5.556	1	5.556	.088	.769
Error(size*const*dir)	2020.237	32	63.132		

Lumbar bending velocity of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	568.480	1	568.480	10.089	.003
Error(size)	1803.075	32	56.346		
const	26.982	1	26.982	.375	.545
Error(const)	2301.303	32	71.916		
dir	8.296	1	8.296	.103	.750
Error(dir)	2580.604	32	80.644		
size * const	9.622	1	9.622	.250	.620
Error(size*const)	1229.613	32	38.425		
size * dir	30.955	1	30.955	.603	.443
Error(size*dir)	1642.745	32	51.336		
const * dir	39.255	1	39.255	.756	.391
Error(const*dir)	1662.635	32	51.957		
size * const * dir	.012	1	.012	.000	.988
Error(size*const*dir)	1827.098	32	57.097		

Lumbar twisting velocity of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	377.285	1	377.285	17.959	.000
Error(size)	672.255	32	21.008		
const	13.455	1	13.455	.257	.615
Error(const)	1672.400	32	52.262		
dir	193.470	1	193.470	3.552	.069
Error(dir)	1742.775	32	54.462		
size * const	2.218	1	2.218	.105	.748
Error(size*const)	674.982	32	21.093		
size * dir	24.121	1	24.121	.514	.479
Error(size*dir)	1501.929	32	46.935		
const * dir	69.444	1	69.444	1.699	.202
Error(const*dir)	1307.961	32	40.874		
size * const * dir	1.069	1	1.069	.024	.878
Error(size*const*dir)	1429.701	32	44.678		

Load moment arm of rear transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	38090.841	1	38090.841	14.390	.001
Error(size)	82056.520	31	2646.985		
const	5612.819	1	5612.819	3.348	.077
Error(const)	51975.142	31	1676.617		
dir	5744.693	1	5744.693	2.396	.132
Error(dir)	74322.369	31	2397.496		
size * const	36.527	1	36.527	.016	.899
Error(size*const)	69649.734	31	2246.766		
size * dir	845.719	1	845.719	.550	.464
Error(size*dir)	47637.942	31	1536.708		
const * dir	2712.457	1	2712.457	1.188	.284
Error(const*dir)	70774.405	31	2283.045		
size * const * dir	246.686	1	246.686	.132	.719
Error(size*const*dir)	58051.275	31	1872.622		

Lumbar flexion angle of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	41.238	1	41.238	1.399	.246
Error(size)	943.144	32	29.473		
const	3745.121	1	3745.121	55.923	.000
Error(const)	2143.008	32	66.969		
dir	276.259	1	276.259	10.787	.002
Error(dir)	819.506	32	25.610		
size * const	59.698	1	59.698	2.188	.149
Error(size*const)	873.154	32	27.286		
size * dir	152.031	1	152.031	7.996	.008
Error(size*dir)	608.440	32	19.014		
const * dir	34.953	1	34.953	2.992	.093
Error(const*dir)	373.795	32	11.681		
size * const * dir	1.128	1	1.128	.073	.789
Error(size*const*dir)	495.848	32	15.495		

Lumbar bending angle of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	33.827	1	33.827	4.777	.036
Error(size)	226.607	32	7.081		
const	783.152	1	783.152	33.558	.000
Error(const)	746.782	32	23.337		
dir	58.149	1	58.149	3.161	.085
Error(dir)	588.625	32	18.395		
size * const	10.125	1	10.125	.659	.423
Error(size*const)	491.399	32	15.356		
size * dir	.123	1	.123	.015	.904
Error(size*dir)	268.781	32	8.399		
const * dir	54.092	1	54.092	4.201	.049
Error(const*dir)	411.992	32	12.875		
size * const * dir	26.410	1	26.410	2.872	.100
Error(size*const*dir)	294.284	32	9.196		

Lumbar twisting angle of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	4.100	1	4.100	2.366	.134
Error(size)	55.446	32	1.733		
const	79.970	1	79.970	13.502	.001
Error(const)	189.531	32	5.923		
dir	.160	1	.160	.054	.817
Error(dir)	94.231	32	2.945		
size * const	1.655	1	1.655	1.097	.303
Error(size*const)	48.262	32	1.508		
size * dir	.170	1	.170	.116	.736
Error(size*dir)	47.056	32	1.471		
const * dir	5.732	1	5.732	3.182	.084
Error(const*dir)	57.649	32	1.802		
size * const * dir	.314	1	.314	.391	.536
Error(size*const*dir)	25.693	32	.803		

Lumbar flexion velocity of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	286.875	1	286.875	7.461	.010
Error(size)	1230.425	32	38.451		
const	53.280	1	53.280	.564	.458
Error(const)	3023.415	32	94.482		
dir	490.909	1	490.909	5.621	.024
Error(dir)	2794.786	32	87.337		
size * const	32.480	1	32.480	.605	.443
Error(size*const)	1719.320	32	53.729		
size * dir	.742	1	.742	.014	.905
Error(size*dir)	1647.888	32	51.496		
const * dir	313.746	1	313.746	5.420	.026
Error(const*dir)	1852.379	32	57.887		
size * const * dir	110.761	1	110.761	2.493	.124
Error(size*const*dir)	1421.789	32	44.431		

Lumbar extension velocity of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	94.788	1	94.788	1.820	.187
Error(size)	1666.791	32	52.087		
const	2021.825	1	2021.825	14.142	.001
Error(const)	4574.792	32	142.962		
dir	63.318	1	63.318	.658	.423
Error(dir)	3078.756	32	96.211		
size * const	174.753	1	174.753	2.655	.113
Error(size*const)	2106.483	32	65.828		
size * dir	77.795	1	77.795	1.569	.219
Error(size*dir)	1586.264	32	49.571		
const * dir	17.665	1	17.665	.190	.666
Error(const*dir)	2981.051	32	93.158		
size * const * dir	6.402	1	6.402	.146	.705
Error(size*const*dir)	1400.410	32	43.763		

Lumbar bending velocity of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	797.269	1	797.269	12.155	.001
Error(size)	2098.896	32	65.590		
const	116.561	1	116.561	.985	.328
Error(const)	3786.167	32	118.318		
dir	54.891	1	54.891	.724	.401
Error(dir)	2426.727	32	75.835		
size * const	8.078	1	8.078	.095	.759
Error(size*const)	2707.514	32	84.610		
size * dir	61.268	1	61.268	1.345	.255
Error(size*dir)	1457.666	32	45.552		
const * dir	271.695	1	271.695	4.512	.041
Error(const*dir)	1927.130	32	60.223		
size * const * dir	300.757	1	300.757	4.342	.045
Error(size*const*dir)	2216.452	32	69.264		

Lumbar twisting velocity of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	96.643	1	96.643	1.960	.171
Error(size)	1577.746	32	49.305		
const	.152	1	.152	.002	.963
Error(const)	2273.075	32	71.034		
dir	18.524	1	18.524	.369	.548
Error(dir)	1606.191	32	50.193		
size * const	16.635	1	16.635	.318	.577
Error(size*const)	1673.642	32	52.301		
size * dir	23.682	1	23.682	.631	.433
Error(size*dir)	1200.614	32	37.519		
const * dir	28.322	1	28.322	.806	.376
Error(const*dir)	1124.568	32	35.143		
size * const * dir	1.357	1	1.357	.053	.820
Error(size*const*dir)	823.954	32	25.749		

Load moment arm of front transferor:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	10437.879	1	10437.879	5.032	.032
Error(size)	66373.934	32	2074.185		
const	863102.367	1	863102.367	176.288	.000
Error(const)	156671.445	32	4895.983		
dir	4.909	1	4.909	.002	.964
Error(dir)	75656.903	32	2364.278		
size * const	14865.004	1	14865.004	8.598	.006
Error(size*const)	55325.309	32	1728.916		
size * dir	5346.000	1	5346.000	3.581	.068
Error(size*dir)	47774.813	32	1492.963		
const * dir	4312.458	1	4312.458	2.678	.112
Error(const*dir)	51538.854	32	1610.589		
size * const * dir	71.095	1	71.095	.095	.760
Error(size*const*dir)	24063.218	32	751.976		

Kinetic Analysis

Compressive force:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	1.138	1	1.138	93.868	.000
Error(size)	.340	28	.012		
const	.002	1	.002	.578	.453
Error(const)	.115	28	.004		
dir	.086	1	.086	14.139	.001
Error(dir)	.171	28	.006		
size * const	.013	1	.013	1.971	.171
Error(size*const)	.190	28	.007		
size * dir	.000	1	.000	.081	.777
Error(size*dir)	.167	28	.006		
const * dir	4.97E-005	1	4.97E-005	.008	.929
Error(const*dir)	.171	28	.006		
size * const * dir	.008	1	.008	3.259	.082
Error(size*const*dir)	.072	28	.003		

Flexion moment:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	.059	1	.059	43.067	.000
Error(size)	.038	28	.001		
const	.001	1	.001	1.512	.229
Error(const)	.011	28	.000		
dir	7.55E-005	1	7.55E-005	.109	.744
Error(dir)	.019	28	.001		
size * const	.001	1	.001	1.717	.201
Error(size*const)	.018	28	.001		
size * dir	.002	1	.002	2.730	.110
Error(size*dir)	.018	28	.001		
const * dir	.000	1	.000	1.166	.289
Error(const*dir)	.011	28	.000		
size * const * dir	.000	1	.000	.240	.628
Error(size*const*dir)	.013	28	.000		

Bending moment:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	.009	1	.009	62.500	.000
Error(size)	.004	28	.000		
const	.001	1	.001	8.306	.008
Error(const)	.004	28	.000		
dir	.000	1	.000	.780	.385
Error(dir)	.011	28	.000		
size * const	5.93E-005	1	5.93E-005	.297	.590
Error(size*const)	.006	28	.000		
size * dir	.000	1	.000	.360	.553
Error(size*dir)	.008	28	.000		
const * dir	2.09E-005	1	2.09E-005	.221	.642
Error(const*dir)	.003	28	9.45E-005		
size * const * dir	4.14E-005	1	4.14E-005	.457	.505
Error(size*const*dir)	.003	28	9.06E-005		

Twisting moment:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	.005	1	.005	13.427	.001
Error(size)	.009	28	.000		
const	7.14E-005	1	7.14E-005	.449	.508
Error(const)	.004	28	.000		
dir	1.82E-005	1	1.82E-005	.050	.825
Error(dir)	.010	28	.000		
size * const	.000	1	.000	.696	.411
Error(size*const)	.005	28	.000		
size * dir	.000	1	.000	1.146	.293
Error(size*dir)	.005	28	.000		
const * dir	.000	1	.000	.841	.367
Error(const*dir)	.006	28	.000		
size * const * dir	6.64E-008	1	6.64E-008	.000	.983
Error(size*const*dir)	.004	28	.000		

Acceleration and RPE Analysis

Rear transferor RPE:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	236.161	1	236.161	117.885	.000
Error(size)	54.089	27	2.003		
const	42.875	1	42.875	9.617	.004
Error(const)	120.375	27	4.458		
dir	3.018	1	3.018	2.886	.101
Error(dir)	28.232	27	1.046		
size * const	.643	1	.643	.462	.503
Error(size*const)	37.607	27	1.393		
size * dir	.071	1	.071	.146	.705
Error(size*dir)	13.179	27	.488		
const * dir	5.786	1	5.786	3.223	.084
Error(const*dir)	48.464	27	1.795		
size * const * dir	.446	1	.446	.391	.537
Error(size*const*dir)	30.804	27	1.141		

Front transferor RPE:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	228.018	1	228.018	128.980	.000
Error(size)	47.732	27	1.768		
const	265.786	1	265.786	47.536	.000
Error(const)	150.964	27	5.591		
dir	2.161	1	2.161	2.194	.150
Error(dir)	26.589	27	.985		
size * const	.875	1	.875	.677	.418
Error(size*const)	34.875	27	1.292		
size * dir	2.571	1	2.571	3.819	.061
Error(size*dir)	18.179	27	.673		
const * dir	3.018	1	3.018	3.584	.069
Error(const*dir)	22.732	27	.842		
size * const * dir	.643	1	.643	1.324	.260
Error(size*const*dir)	13.107	27	.485		

Vertical acceleration:

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
size	.103	1	.103	1.319	.261
Error(size)	2.025	26	.078		
const	.008	1	.008	.196	.661
Error(const)	1.021	26	.039		
dir	.951	1	.951	11.545	.002
Error(dir)	2.141	26	.082		
size * const	.004	1	.004	.118	.734
Error(size*const)	.963	26	.037		
size * dir	.260	1	.260	5.499	.027
Error(size*dir)	1.228	26	.047		
const * dir	.124	1	.124	1.668	.208
Error(const*dir)	1.929	26	.074		
size * const * dir	.105	1	.105	1.820	.189
Error(size*const*dir)	1.505	26	.058		