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# Designing ergonomic interventions for EMS workers, Part I: Transporting patients down the stairs

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**Abstract**

The objective of the current work was to test ergonomic interventions aimed at reducing the magnitude of trunk muscle exertions in firefighters/paramedics (FFPs) providing emergency medical services (EMS) when transporting patients down the stairs. The interventions, developed using focus groups, were a footstrap to prevent the patient from sliding down on the backboard, a change in the handle configuration on the stairchair, and 2 devices, the ''backboard wheeler'' and a tank tread-like device (descent control system, DCS) for a stretcher, that change the backboard and stretcher carrying tasks into rolling and sliding tasks. Eleven two-person teams transported a 75 kg dummy with each intervention and its corresponding control condition down a flight of steps. Surface electromyographic (EMG) data were collected from 8 trunk muscles from each participant. Results showed that the backboard footstrap reduced the erector spinae (ERS) activity for the FFP in the ''leader'' role by 15 percent, on average. The change in handle configuration on the stairchair had no effect on the variables measured. The backboard wheeler reduced the ERS activity bilaterally in the FFP in the leader role and unilaterally for the FFP in the ''follower'' role, by 28 and 24 percent, respectively. The DCS reduced the 90th percentile ERS activity for both FFPs from 26 to 16 percent MVC, but increased the latissimus dorsi activity in the follower from 11 to 15 percent MVC. The DCS was the only intervention tested that resulted in a reduced rating of perceived exertion relative to the corresponding control condition. In summary, the hypotheses that the proposed interventions could reduce trunk muscle loading were supported for 3 of the 4 transport interventions tested.


*Keywords:* LBD; EMS; Intervention

## 1. Introduction

Musculoskeletal injuries consistently account for about half of all injuries to firefighters/paramedics (FFPs) engaged in emergency medical service (EMS) operations as well as to EMS workers in the private sector (Karter and Molis, 2004; Reichard and Jackson, 2004). These injuries result in lost work time, permanent disability, and high worker compensation costs (Karter and Molis, 2004; Walton et al., 2003). In a recent analysis of 1343 firefighter worker compensation claims, the per-claim average worker's compensation cost for sprain/strain injuries was over 50 percent greater than for claims overall (Walton et al., 2003). The back was the primary body part affected.

As the prevalence of obesity in the general population increases so does the risk for injury to the EMS workers who are responsible for transporting these heavier patients (Centers for Disease Control and Prevention (CDC) and National Center for Health Statistics (NCHS), 2004). The tasks performed during EMS runs that place these workers at risk for these serious sprain and strain-type injuries include lifting and carrying patients for transport to the hospital (Lavender et al., 2000a, b). In fact, a recent study reported that the leading cause of injury to EMS workers is the patient (Reichard and Jackson, 2004). Research shows

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that current methods for transporting patients need improvement in order to better protect the EMS workers from injury (Lavender et al., 2000b).

The purpose of this work was to evaluate ergonomic interventions aimed at reducing biomechanical loads on the back as patients are transported down the stairs during EMS operations. Our previous findings, based on interviews and surveys with FFPs, indicated that stair decent tasks are both strenuous and frequently performed (Conrad et al., 2000). This was confirmed in biomechanical studies for both members of the two-person team that routinely perform this task (Lavender et al., 2000a, b). FFPs participating in our focus groups reported that they frequently transport patients from upstairs bedrooms to street level using a stairchair, a backboard, or a stretcher. The selection of the specific piece of equipment is based upon the health status of the patient, space constraints, equipment weight, and individual preferences. Stairchairs, for example, are frequently used where there are tight corners, although relative to a backboard or the stretcher, require an additional transfer prior to being placed in the ambulance. Backboards are used for individuals who cannot be transported in a seated posture, and while they can be placed directly on the stretcher in the ambulance, may be difficult to maneuver through a 90° or 180° turn on a staircase. The same limitation applies to transporting patients directly on a stretcher; however, there is also the increased weight of the stretcher to consider which often exceeds 36 kg.

Several ergonomic interventions were developed through a series of focus group discussions with FFPs. Three groups, each with between 8 and 10 FFPs met twice to brainstorm and refine intervention ideas. An industrial designer sketched drawings of the concepts suggested by the participants. During these discussions, the ergonomic challenges of using existing equipment were discussed and potential solutions to the ergonomic problems were identified. These included modifications to existing equipment and the development of new approaches. This paper describes a series of interventions aimed at reducing the trunk muscle exertions during stair-transport tasks. Each of the interventions focused on assisting a two-person team. Within each team we identified 2 roles. The FFP descending the stairs first, usually walking backwards, we refer to as being in the "leader" role. The "follower" role was played by the FFP trailing behind the patient and the transport apparatus.

The first intervention was a footstrap that attached to the backboard to prevent the patient from sliding toward the leader. When a patient slides so that his/her feet extend beyond the backboard, the leader ends up holding the backboard further from the body thus increasing the biomechanical load on the leader (Fig. 1a). The second intervention for the stair descent tasks was a device that allowed the backboard to be rolled down the stairs on a handtruck-type device, thereby changing the carrying task into a push and pull task (Fig. 1b). The third intervention was a set of "tank treads" that allow a stretcher to be rolled down the stairs by bridging the steps, thus changing the task from a lift and carrying task to a push and pull task, for the leader and follower, respectively (Fig. 1c). The fourth intervention was a change in the handle configuration of the stairchair (Fig. 1d). Many of the FFPs had indicated in the focus groups that the handles available on their stair chair resulted in their hands being held further from the body to eliminate contact with the patient's head, thus increasing the moment arm between the hand positions and the spine. In each case the hypothesis being tested is that the intervention results in lower electromyographic (EMG) activity than that found with the current equipment. EMG was selected as the primary dependent measure as it provides an objective measure of the exertion magnitude. The aim of the interventions developed through this process was to reduce the magnitude of the exertions required by the major trunk muscles that support, move, and stabilize the spine.

## 2. Methods

### 2.1. Subjects

Eleven two-person teams of FFPs were recruited from fire departments in the Chicago suburbs. All participants were employed fulltime as FFPs at their respective departments. Assignment to a specific team was based on availability. Similar to actual work situations, the team members designated which FFP was the leader and which FFP was the follower. Ten of the teams were comprised of 2 males. One team was comprised of 2 females. Each participant signed an informed consent prior to participating in the study.

The mean age, height, and weight of the subjects was 37 years (28–51 years), 1.80 m (1.63–1.89 m), and 96 kg (70–123 kg), respectively. On average the FFPs had 12 years of experience in the EMS component of the job (<1–25 years). None were experiencing low back pain at the time of the study.

### 2.2. Experimental design

The experiment examined the 4 specific interventions using a repeated measures design. Specifically, each intervention was compared with the existing method and equipment (control condition). Thus, this investigation required each two-person team to perform several stair descent tasks. The sequencing of the experiments was randomized for each two-person team as was the sequencing of the experimental versus control condition. Each intervention was evaluated across the series of subtasks we had identified in our previous investigations (Lavender et al., 2000a, b), for example, level carry, beginning the stairs ("begin stairs"), continuing down the stairs ("mid-stair"), and making a 90°turn through a landing ("landing").


Fig. 1. The 4 interventions evaluated in this paper. Clockwise from the top left: the backboard footstrap (a), the backboard wheeler (b), the decent control system (DCS) (c), and the extended handle stair chair (d).

The primary dependent measures were the surface EMG signals from 8 trunk muscles. The muscles sampled included the left and right latissimus dorsi (LATL and LATR), erector spinae (ERSL and ERSR), external oblique (EXOL and EXOR) and rectus abdominus (RABL and RABR). After normalizing each muscle relative to a maximal contraction, the mean and 90th percentile level of each muscle's activity was extracted for analysis.

### 2.3. Apparatus and instrumentation

To evaluate the stairchair handle configurations we purchased a specialized configuration of a new stairchair model from Stryker™ (Model 6250), which weighs 10.5 kg. On this chair the conventional handles were 33 cm above the seat pan. The extendable handle could be positioned 71 or 97 cm above the seat pan. Shorter individuals elected to use the extended handle at the lower setting. In both cases the handle was positioned above the patient's head.

This study used the tapered Najo Backboard by Ferno Washington™, which weighed 7.7 kg. The footstrap was comprised of a belt and a 90° angle bracket that was placed beneath the patient's feet and held the belt so it would not slide toward the board, which would have allowed it to slip out from under the patient's feet. The belt and the bracket added 0.6 kg to the total weight of the backboard.

The stretcher used in the evaluation of the tank tread devices was manufactured by Stryker (MX3-Pro™) which weighs 37 kg. The tank tread devices were a prototype of the descent control system (DCS) developed by Paramed Systems. The prototype DCS version added an additional 6.4 kg to the stretcher. Because these devices could be easily disengaged but not easily removed from the stretcher, the control condition was conducted with the devices still attached to the stretcher. The passive braking system within the DCS was adjusted for maximum resistance. The active handbrake system was not used.

The backboard wheeler was fabricated at the scientific instrument shop at the University of Illinois at Chicago. Due to some significant changes made to the initial prototype device, data are only available for the last 5 teams. The final prototype device weighed 19.5 kg. The handle for the leader varied between chest and elbow level as the wheeler descended each step. The handles for the

follower varied between elbow and mid-thigh levels while descending the stairs.

For each transport task a 75 kg dummy was moved. This kind of dummy is used by the fire service for training purposes. With the exception of the footstrap comparison, the dummy's position was the same for the control and intervention conditions. For the "control" condition in the absence of the footstrap intervention, the dummy was shifted such that the feet extended 15.2 cm beyond the end of the backboard to simulate the slippage problem.

The FFPs were each instrumented with a lumbar motion monitor (LMM) to quantify trunk postures. Each FFP was connected to an 8-channel telemetered EMG system (Noroxan). The RMS output from the Noroxan amplifiers was sampled along with the kinematic data using InnSport™ data collection software at 120 Hz for 6 s. Heart rate data were obtained using Polar™ Heart rate monitors. Subjects read the heart rate off the wristwatch display when polled by one of the investigators before and after the performance of each stair descent task. The measure used in the analysis was the number of beats per minute above the resting rate measured at the beginning of each experiment.

The tests were conducted on a staircase that was 91 cm wide between the railings. The subjects descended 15 steps before coming to a landing that was 91 × 137 cm where a 90° turn to the right (facing forwards) was required. The rise and run dimensions of the steps were 18.4 and 26.7 cm, respectively. Nearly all FFPs participating in the focus groups indicated that they would not use a stretcher if there was a landing involved. Thus, all stretcher tests terminated when the leader reached the landing. For the other devices, the FFPs were asked to complete the turn and take at least 1 step on the next flight of stairs.

### 2.4. Task protocol

After reading and signing the informed consent documents and watching an instructional video that demonstrated how to use the experimental interventions, each member of the team was instrumented with disposable surface electrodes (Cleartrace™ 1700-030) at standard muscle sites as described by Marras (1987). For the erector spinae the electrodes were positioned approximately 5 cm lateral from the midline at the L3 level. The latissiums dorsi muscles were positioned at the T7 level over the belly of the muscle, approximately 13–15 cm lateral from the midline. The external oblique electrodes were positioned at the level of the umbilicus and centered approximately halfway between the iliac crest and the anterior superior iliac spine at an angle of 45°. The rectus abdominus muscles were placed just above the umbilicus approximately 2.5 cm lateral from the midline.

Prior to conducting the study the subjects performed 2 types of maximal voluntary exertions to obtain maximal EMG signals for normalization purposes. In the trunk extension exertions subjects pulled up on a handle positioned at approximately mid-thigh level. This resulted in a modest degree of spine flexion (~20°), similar to what we expected during the more strenuous points in the task. This task was repeated until maximal EMG values were obtained from both the erector spinae and latissimus dorsi muscles. The trunk flexion exertions were completed by connecting a cable between a chest harness and a reference frame apparatus (Lavender et al., 1992). These provided maximal signals from both sets of abdominal muscles. In reality, because of the weight of the dummy, often these "maximal" values were exceeded during the testing protocol. When this occurred, the maximal values were replaced with the highest observed EMG value from the experimental trials.

Before testing the more novel interventions, for example, the DCS and the backboard wheeler, the FFPs were encouraged to take a practice run down the stairs before data were collected for the single trial representing each experimental condition. Practice was limited, however, to avoid fatiguing the subjects with multiple trials. Once the stair descent was completed, the FFPs were asked to leave the dummy and tested apparatus on the landing. Members of our research team returned the dummy to the top of the stairs in preparation for the next experimental condition. During each task, the FFPs were encouraged to verbally communicate with each other as they typically would on the job. At the completion of each exertion the subjects were polled for their current heart rate and to provide a rating of perceived exertion using the Borg CR10 (Borg, 1998) between 1 (easiest) and 10 (hardest). The stair descent tasks that included the landing lasted between 20 and 25 s. The straight stair descent task without the landing (stretcher transport) lasted approximately 15 s.

### 2.5. Data analysis

The mean and 90th percentile EMG readings from each subtask were normalized to relative maximum and resting levels. The analysis of the LMM data focused on trunk posture, hence the maximal postural deviation during each subtask was extracted from the data stream from each FFP's LMM. Both the EMG and the LMM data were first analyzed using multivariate analyses of variance (MANOVA) in the SAS software. An alpha level of 0.05 was used for all multivariate analyses except for those used in the analysis of the backboard wheeler. Because we had only 5 teams use the final version of the backboard wheeler, we relaxed the multivariate significance level to 0.10 in recognition of the reduced power. Where multivariate analyses were significant, univariate ANOVAs were run to identify the individual muscles or postures responsible for their respective multivariate effects. The heart rate data were pooled across both team members for each experimental condition and analyzed using a paired $t$-test. The ratings of perceived exertion were analyzed for each intervention by role using a paired $t$-test.

## 3. Results

### 3.1. Backboard footstrap

The MANOVA for the normalized EMG data indicated there were significant changes in both the mean and 90th percentile EMG data for the leader when using the footstrap on the backboard (Table 1). There were no changes in the follower's muscle activation levels due to the footstrap. For the leader, both erector spinae muscles showed reductions in average activity and 90th percentile activation levels with the footstrap (Fig. 2). While the muscle activity was greatest during the landing subtask, the footstrap's effect on muscle recruitment was consistent across the subtasks. Overall, the ERSR and ERSL showed a decrease in mean activity of 11 and 22 percent, and a decrease in 90th percentile activity of 10 and 19 percent, respectively. Although the activation level was low with both the footstrap and in the control condition, the RABL showed a significant decrease in average activation level with the footstrap. Our measure of total workload, heart rate, was essentially unchanged due to the footstrap intervention (Fig. 3) as were the ratings of perceived exertion (Table 2).

The spine postures were not different due to the intervention (Table 3). Irrespective of the intervention, the posture when carrying the backboard was upright, with little side bending (~6°) and a small amount of twisting

Table 1
*P*-values from the multivariate and univariate tests performed on the average muscle response over the sampling period (mean) and 90th percentile (90%ile) muscle response data

| Intervention | Analysis | Leader | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | | MANOVA | LATR | LATL | ERSR | ERSL | RABR | RABL | EXOR | EXOL |
| Backboard footstrap | Mean | 0.0021 | ns | ns | 0.0102 | 0.0003 | ns | 0.0394 | ns | ns |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | 0.0032 | ns | ns | 0.0075 | 0.0005 | ns | ns | ns | ns |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| Backboard wheeler | Mean | 0.028 | ns | ns | 0.0005 | 0.0169 | ns | ns | ns | ns |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | 0.1085 | ns | ns | 0.0013 | ns | ns | ns | ns | ns |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| Extended handle Stairchair | Mean | ns | . | . | . | . | . | . | . | . |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | ns | . | . | . | . | . | . | . | . |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| DCS stretcher | Mean | 0.0013 | 0.043 | ns | 0.0002 | 0.0095 | ns | ns | ns | 0.0012 |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | 0.0073 | 0.0174 | ns | 0.0003 | 0.0106 | ns | 0.0074 | ns | 0.0074 |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | | Follower | | | | | | | | |
| Backboard footstrap | Mean | ns | . | . | . | . | . | . | . | . |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | ns | . | . | . | . | . | . | . | . |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| Backboard wheeler | Mean | 0.0726 | ns | ns | ns | 0.0342 | | 0.0184 | | 0.0566 |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | 0.0871 | ns | ns | ns | 0.0721 | ns | ns | ns | ns |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| Extended handle Stairchair | Mean | ns | . | . | . | . | . | . | . | . |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | ns | . | . | . | . | . | . | . | . |
| | Interaction | 0.0605 | 0.0277 | 0.036 | ns | 0.0037 | ns | ns | 0.067 | ns |
| DCS stretcher | Mean | 0.0002 | ns | 0.0126 | 0.0001 | 0.0001 | ns | ns | 0.0001 | |
| | Interaction | ns | . | . | . | . | . | . | . | . |
| | 90%ile | 0.0003 | ns | ns | 0.0001 | 0.0001 | 0.0559 | ns | 0.0007 | 0.0339 |
| | Interaction | ns | . | . | . | . | . | . | . | . |

Mean = NEMG data averaged across sampling period.
Interaction = intervention*subtask interaction.
ns = non significant (>.10 for multivariate, >.05 univariate).
. = not analyzed due to non-significant multivariate effect.

(~8°). However, when the FFPs were carrying the backboard through the landing there was more forward bending on both the part of the leader (average = 22°) and the follower (average = 15°).

### 3.2. Extended handle stairchair

The stairchair intervention was primarily developed to allow the follower better access to the handles. Hence, the MANOVA conducted on the 90th percentile posterior trunk muscle responses indicated that there were no significant changes in the muscle recruitment on the part of the leader (Table 1). For the follower, there were significant changes in the 90th percentile and mean responses of selected muscles that were dependent upon the subtask performed. Closer analysis of this interaction effect found that both the latissimus dorsi muscles and the ERSL showed significantly greater activity as the stairchair with the extended handle was carried through the 90° turn on the landing (Fig. 4). These results are consistent with the heart rate data which showed a significant increase, on average 4.8 beats/min, while using the extended handle stairchair (Fig. 3). The mean muscle response was not affected by the handle intervention thereby indicating that the changes seen in the 90th percentile values were short variations in the peak activity as the extended handle chair was swiveled through the landing. The ratings of perceived exertion did not change for either role (Table 2).

As with the backboard carries, the spine postures were generally upright and with little lateral bending or twisting. The only change in the spine posture associated with the extended handle was a 2.0° decrease in the twisting posture observed in the follower ($p = .001$). Overall, there was between 3° and 5° more twisting on the part of the leader as the stairs were initiated and descended than observed for

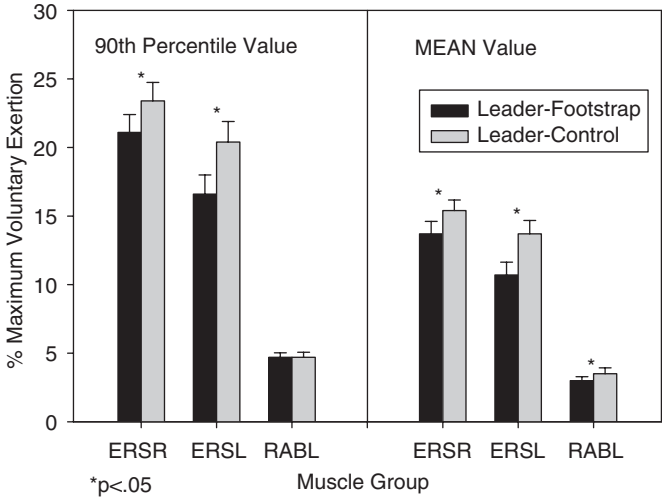


Fig. 2. The effects of the backboard footstrap on the leader's average and 90th percentile erector spinae and the left rectus abdominus muscle activations. Error bars represent the standard error of the mean.

Table 2
The mean ratings of perceived exertion (RPE) obtained from the leader and follower following each experimental condition

| Experimental condition | Leader | Follower |
|---|---|---|
| Stretcher | 5.5 | 5.0 |
| DCS | 3.3 | 2.3 |
| Conventional SC | 3.1 | 2.5 |
| Extended handle SC | 3.3 | 2.9 |
| Backboard wheeler | 3.0 | 3.6 |
| Backboard footstrap | 4.4 | 3.6 |
| Backboard control | 4.6 | 3.5 |

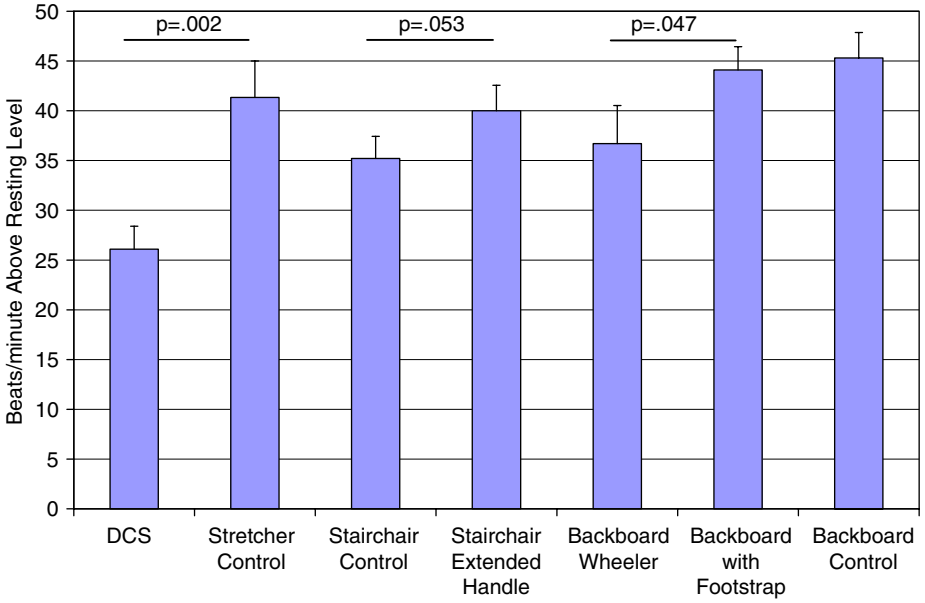


Fig. 3. The incremental heart rate obtained at the completion of each transport as a function of the equipment used. Error bars represent the standard error of the mean.

Table 3
*P*-values from the multivariate and univariate tests performed on the postural data obtained with the LMM

| Intervention | Role | Effect | MANOVA | Forward bend | Lateral bend | Twist |
|---|---|---|---|---|---|---|
| Backboard footstrap | Leader | Intervention | ns | . | . | . |
| | | Interaction | ns | . | . | . |
| | Follower | Intervention | ns | . | . | . |
| | | Interaction | ns | . | . | . |
| Backboard wheeler | Leader | Intervention | 0.0009 | 0.0001 | ns | ns |
| | | Interaction | 0.0519 | 0.0071 | ns | ns |
| | Follower | Intervention | 0.0001 | 0.0035 | 0.0006 | ns |
| | | Interaction | ns | . | . | . |
| Extended handle Stairchair | Leader | Intervention | ns | . | . | . |
| | | Interaction | ns | . | . | . |
| | Follower | Intervention | 0.0168 | ns | ns | 0.0012 |
| | | Interaction | ns | . | . | |
| DCS stretcher | Leader | Intervention | 0.0001 | 0.0226 | 0.0001 | ns |
| | | Interaction | ns | . | . | . |
| | Follower | Intervention | 0.0213 | ns | 0.0037 | ns |
| | | Interaction | ns | . | . | . |

Interaction = intervention*task interaction.

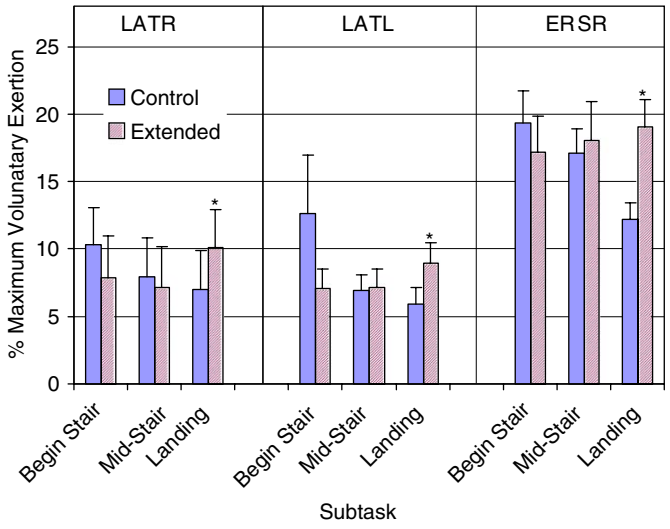


Fig. 4. Averaged 90th percentile normalized EMG values for the 3 muscles in the follower that showed a significant handle by subtask interaction when carrying the patient on the stairchair. Error bars represent the standard error of the mean.

the follower, probably due to the requirement that leader walk backwards down the stairs ($p<.001$).

### 3.3. Stretcher equipped with the DCS

The DCS significantly decreased both the average and the 90th percentile activation levels of the erector spinae muscles (Table 1, $p<.01$). The latissimus dorsi muscles also showed reduced average and 90th percentile activity for the leader; however, the DCS led to an increase in the average LATL activation in the follower. The effect of the intervention was consistent across the begin and mid-stair tasks, hence averaged 90th percentile activation levels measured during the DCS and the control (stretcher lift and carry) conditions are shown in Fig. 5. The mean responses over the 6-s sampling period showed a similar trend due to the intervention. For example, the 90th percentile erector spinae response (averaged bilaterally) for the follower decreased 37 percent while the mean response decreased by 35 percent. There was also some reduction in the anterior muscle activities for both FFPs; however, there were differences in the specific muscles changing their activation levels across the 2 roles.

The DCS resulted in the most notable decrease in heart rate across the interventions tested. Across both team members the heart rate dropped 15 beats/min when using the DCS as compared with the stretcher carry. The ratings of perceived exertion dropped significantly by 2.2 points for the leader and 2.7 points for the follower (Table 2).

On average, when using the DCS, both the leader and the follower adopted spine postures with 3° and 6° more lateral flexion, respectively, than when carrying a stretcher (Table 3, $p<.05$). The leaders' forward bending increased by nearly 7° when using the DCS to a value of 16° ($p<.05$) compared to an average of 9° when carrying the stretcher. The follower's forward bending did not differ across intervention conditions and averaged 8.3°.

### 3.4. Backboard wheeler

In the analysis of the backboard wheeler, the wheeler was compared to the backboard carry with the footstrap intervention as this was the condition in which the dummy was secured in the correct position on the backboard. Due to some design modifications made midway through the testing process, data were only available from the last 5 teams. The backboard wheeler had a more pronounced effect on the leader as the nature of the task changed from

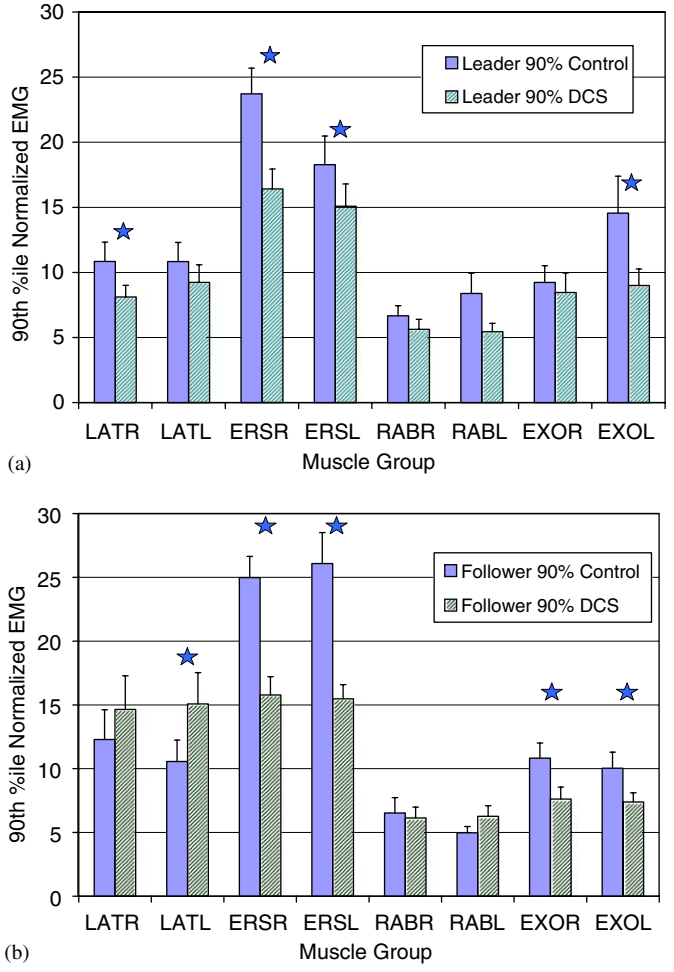


Fig. 5. Averaged 90th percentile normalized EMG values for the leader and the follower when descending the stairs with the patient using the DCS versus when being carried on a stretcher. Error bars represent the standard error of the mean.

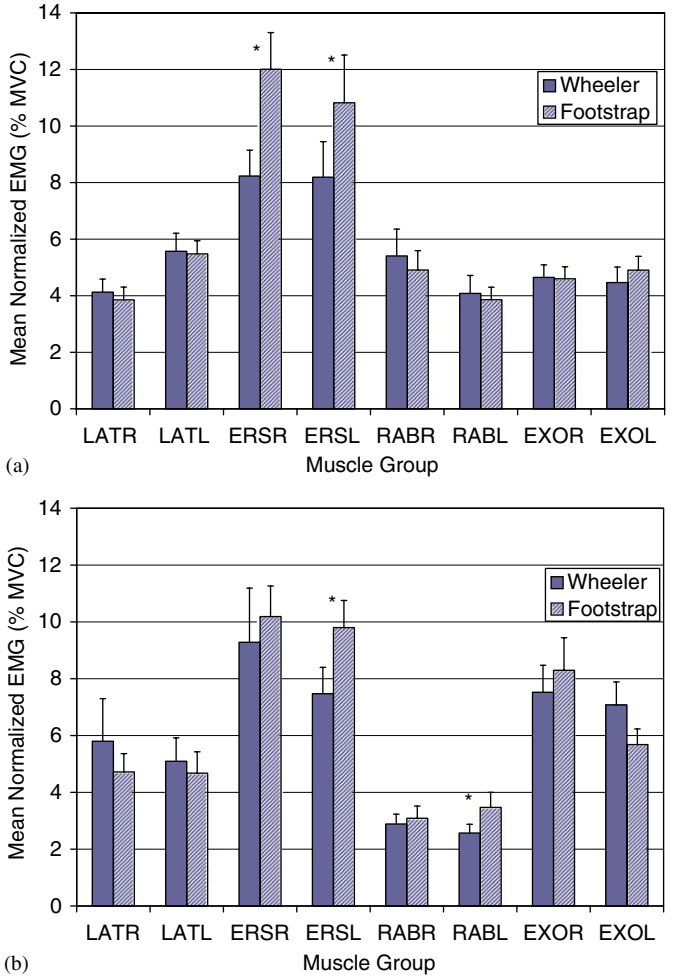


Fig. 6. Time-averaged normalized EMG for the leader (a) and the follower (b) when descending the stairs with the patient on the backboard wheeler versus when being carried on a backboard. These results were based on a smaller sample ($n = 5$). Error bars represent the standard error of the mean.

carrying to controlled lowering as the person in this role typically eased the transition from one step to the next. For both the leader and the follower the changes were more robust when looking at the time averaged normalized EMG signal than when looking at the 90th percentile values. Fig. 6 shows the mean normalized EMG averaged across the 6-s sampling windows and across the 4 subtasks (level carry, begin stairs, mid-stairs, 90° turn on landing). At the multivariate level, the test of the intervention main effect showed low $p$-values only for the leader ($p = .028$ for the leader, $p = .0726$ for the follower). The significant ($p < .05$) univariate comparisons are indicated in Fig. 6 for both roles. The strongest findings were the bilateral reductions of the erector spinae response observed in the leader, and the ERSL and RABL response of the follower. Table 2 shows that there was no change in the ratings of perceived exertion made by the FFPs in the follower role; however, there was a trend towards a lower perceived exertion rating by those in the leader role ($p = .108$, $n = 5$).

Table 4
The maximum postural deviation observed averaged across the 5 leaders when carrying the patient on a backboard versus when using the backboard wheeler for each of the tasks

| Task | Backboard carry | | Backboard wheeler | |
|---|---|---|---|---|
| | Mean | Std. dev. | Mean | Std. dev. |
| Level ground | 1.7 | 9.6 | 11.8 | 12.6 |
| Begin stairs | 9.8 | 22.0 | 0.1 | 10.2 |
| Mid-stairs | 5.1 | 13.4 | 3.3 | 10.5 |
| Landing | 12.7 | 11.7 | 22.9 | 17.7 |

On average the backboard wheeler increased the lateral bending from 4° to 8° for both the leader and the follower, although the change was only significant for the follower. The amount of forward bending on the part of the leader was dependent upon the subtask as well as the intervention (Table 3). As shown in Table 4, there was increased forward bending by the leader on level ground and in the

landing when using the wheeler. Often the leader FFP would bend forward and lift the front end of the wheeler to facilitate the 90° turn. However, the leader was also bent forward when carrying the backboard through the landing in an effort to effectively shorten the backboard and reduce the space required to make the turn. During the level carry subtask, there was more flexion on the leader's part as this person sometimes held the handle and assisted in the steering, although there was generally little external load applied. Once on the steps, the flexion angles were similar. When the sagittal velocity data were examined it was found that the speed of the forward bending motion was very similar for both roles and that the velocity increased by a factor of 1.8, from 18 to 33°/s, with the backboard wheeler. While 33°/s is still relatively slow, it is suggestive of the repeated bending motion as the wheeler was lowered to each successive step.

## 4. Discussion

Overall, the data indicate that 3 of the 4 interventions were successful in reducing back muscle use relative to approaches currently used. While anterior muscles were measured and some significant changes observed, much more activity was observed in the posterior muscles, specifically, the erector spinae muscles. These muscles, while supporting and stabilizing the torso, provide much of the force serving to compress the spine during these heavy patient transport tasks. Thus, decrements in the erector spinae activity are believed to represent the most important findings presented. There were also subtle changes in the trunk kinematics due to the interventions. However, most of the postures were close to upright making the functional significance of these changes, which could be distributed across the lumbar and lower thoracic spine, questionable. The heart rate data support the hypothesized changes in workload.

It should be noted that these tests were conducted with little practice prior to using the interventions. While increased practice may have allowed us to see the true potential of each intervention, this was difficult to implement given the already large time and expense commitment involved with recruiting professional FFPs. The additional practice would have been most beneficial with the DCS and the backboard wheeler as these interventions were the most different from current work practices. On the other hand, by testing with little practice, we have observed a beneficial effect that would be potentially available from the point at which these interventions are introduced to the workforce. It is anticipated that greater experience would likely translate to even larger reductions in muscle recruitment as co-contraction is lessened with the development of behavioral skills.

In this study we chose to analyze both the mean and the 90th percentile normalized EMG values. The mean was chosen to provide an indicator of the overall recruitment level that was required throughout each subtask. The 90th percentile value was selected as a more robust estimation of the peak value that occurs within each task. While historically maximum values have been used, these are more susceptible to movement artifacts, temporary increases in pressure on the electrodes, or in our case radio transmission artifacts. Generally the 90th percentile and the mean responses showed good agreement for each of the interventions tested. The data from the backboard wheeler tests, however, showed more separation between the means than the 90th percentile values, particularly for the FFP in the follower role. This suggests that overall the wheeler reduced the activity, yet the peaks in activity tended to be similar to those experienced while carrying the board.

### 4.1. Backboard footstrap

The footstrap is a relatively simple intervention. The advantage of the footrest device, which is part of our footstrap, is that it provides a path for the footstrap that prevents the strap from sliding towards a less effective position at the patient's heels. The primary biomechanical benefit of this intervention was to reduce the moment arm between the hand coupling and the leader's spine. By keeping the patient from sliding down the board, the leader is able to hold the board right next to the body thereby minimizing the moment generated about the spine. While the moment arm was not actually measured in this study, the reduction in erector spinae EMG activity was consistent with the reduction in the net external moment that would result from the reduction in the moment arm. There essentially was no effect for the follower. The value of this intervention is that would be inexpensive to produce and could likely be implemented by most FFPs with existing materials.

### 4.2. Extended handle stairchair

The goal of extending the handles on the stair chair was to minimize the distance between the handles and the torso for the follower FFP. Often the patient's head prevents the stairchair from being held right next to the FFP's body. While, extending the handle appeared to have moved the patients head away from the FFP's body, for shorter individuals it required more shoulder effort to hold the stairchair high enough so that it could clear the stairs. The other observation was that the "normal" handles on the Stryker stairchair tested were relatively long and therefore may have not represented the conditions described in the initial focus groups. In fact, the handles used in the control condition may have allowed the FFPs to keep their hands more or less along side the body, thereby minimizing the load moment created by the chair and removing the hypothesized benefit of the intervention.

### 4.3. Stretcher equipped with the DCS

Even though most of the FFPs tested indicated they would not carry a stretcher upstairs to retrieve a patient,

most found the DCS relatively easy to use. The erector spinae EMG data strongly support its use when compared with carrying the stretcher. In fact, across all the methods tested for transporting patients down the stairs the DCS resulted in the lowest erector spinae muscle recruitments for both the beginning and mid-stair tasks across both the leader and follower roles. It should be noted that there was an increase in the activity of the latissimus dorsi in the follower. This is consistent with the change in direction of the external force acting on the follower FFP. When carrying the stretcher the force direction is vertical, resulting in a static lifting task. The force vector with the DCS matched the slope of the stairs, thus containing a much larger horizontal component. Hence, in addition to the upwards exertion, the rearward component of the exertion would be expected to increase latissimus dorsi recruitment (Lavender et al., 1998). It should also be noted that the resulting latissimus dorsi recruitment levels were still relatively low compared with the erector spinae recruitment levels observed under the control condition. In addition, use of the DCS would also eliminate an additional patient chair to stretcher transfer at the completion of the stairchair transport task.

### 4.4. Backboard wheeler

In response to verbal feedback from the first 6 teams, it became clear that the rear axle needed to be shifted so as to change the center of gravity of the patient/wheeler system. This change substantially reduced the hand force required by the follower, but also eliminated the data from the initial 6 teams in the analysis. Unfortunately, this meant that there was a relatively small sample size available for evaluating this intervention ($n = 5$). As previously mentioned, the test was conducted using the backboard equipped with the footstrap as the comparison. As a result, we believe this was a conservative test of the backboard wheeler given that the footstrap prevented the patient from sliding toward the leader as commonly reported by our subjects. However, despite the small sample size, the significant bilateral reduction in erector spinae muscle recruitments on the part of the leader, and the smaller reductions observed on the part of the follower, suggest that the device holds promise based on the limited experimental data obtained.

One limitation of this work is that while individual muscle recruitments were analyzed, the collective effect on the biomechanical loading of the spine was not assessed. However, the primary determinant of the aggregate compression and shear forces are the individual muscle loadings. If these are reduced through the interventions then we can be reasonably confident that the interventions would reduce the biomechanical loads experienced by the inter-vertebral discs. A second limitation was the potential error in the normalized EMG given that the maximal extension EMG was obtained in a single moderately flexed posture (20°). Clearly this single posture is a compromise as some tasks were performed with very little flexion and some with more. While this may affect the absolute normalized EMG level, we believe that given the relatively small changes in postural conditions due to the interventions tested, the relative comparisons between normalized EMG values across conditions are valid for evaluating the interventions.

## 5. Conclusions

The hypotheses that the proposed interventions could reduce muscle recruitments were supported for 3 of the transport interventions tested. Only the extended handle on the stairchair failed to reduce back muscle recruitment. While the analyses presented in this paper stopped short of computing spine loads, we believe that the reduced muscle exertions, as evidenced by the observed changes in the EMG data, support the use of these interventions in that they have the potential to reduce over-exertion-type injuries.

The footstrap is easily implemented with materials most EMS crews have on hand and will add little time to the patient preparation. Preventing the patient from sliding down the board, in addition to helping out the FFP in the leader role will likely make the transfer less uncomfortable for the patient. The DCS and potentially the backboard wheeler, in transforming the carrying task to a push pull task, will likely be the most beneficial approaches to preventing back injuries associated with the transport of patients.

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