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Telehealth Monitor to Measure Physical Activity and Pressure Relief Maneuver Performance in Wheelchair Users

Ariel V. Dowling, PhD[~], Valerie Eberly, PT^{*}, Somboon Maneekobkunwong, MSME^{*}, Sara J. Mulroy, PhD, PT^{*}, Philip S. Requejo, PhD^{*}, and Joseph T. Gwin, PhD[~]

[~]BioSensics LLC, Cambridge, MA

^{*}Pathokinesiology Laboratory, Rancho Los Amigos National Rehabilitation Center, Downey, CA

Abstract

The purpose of this study was to demonstrate the feasibility of a device for monitoring pressure relief maneuvers and physical activity for wheelchair users. The device counts the number of wheel-pushes based on wheelchair acceleration and measures pressure relief maneuvers using a seat-sensor consisting of three force sensing resistors.

To establish the feasibility of the seat-sensor for the detection of pressure relief maneuvers, ten wheelchair users and ten non-disabled controls completed a series of wheelchair depression raises, forward trunk leans, and lateral trunk leans. The seat-sensor was placed underneath the user's own seat cushion. To establish the feasibility of wheel-push counting, ten full-time wheelchair users navigated a flat 50 m outdoor track and a 100m outdoor obstacle course during self-propulsion (e.g., wheel-pushes) and during assisted-propulsion (e.g., someone else pushing the wheelchair, no wheel-pushes).

Of the 240 performed pressure relief, 225 were properly classified by the seat-sensor (accuracy: 94%, sensitivity: 96%, specificity: 80%). All four types of maneuvers had accuracy above 94%. Sensitivity was highest for depression raises (98%) and lowest for front lean maneuvers (80%). The wheelchair activity monitor measured 2112 pushes during the self-propulsion trials compared to 2162 pushes measured with the instrumented push-rim (97.7%). During assisted-propulsion trials there were 477 incorrectly identified pushes (8.0 per trial).

Keywords

Wheelchair; activity monitor; pressure relief; accelerometer

Introduction

People with disabilities experience costly and preventable secondary health conditions (Havercamp, Scandlin, & Roth, 2004). Higher proportions of people with disabilities indicate that their health is fair or poor when compared to healthy adults, and those with the severest disabilities report the worst health (Drum et al., 2009). Therefore, services and

technologies that provide health education and promote self-management are a critical piece of comprehensive systems of care for people with disabilities (Campbell, Sheets, & Strong, 1999).

A significant contributing factor to the development of secondary conditions among individuals with disabilities is inactivity (Campbell et al., 1999; Krause & Saunders, 2011; Walter et al., 2002). Many people who use a wheelchair have inactive lifestyles (Bussmann, Ebner-Priemer, & Fahrenberg, 2009; Coulter, Dall, Rochester, Hasler, & Granat, 2011; Noreau & Fougereyrollas, 2000). This can cause secondary complications such as obesity, diabetes, osteoporosis, and cardiovascular comorbidities (Sullivan, Morrato, Ghushchyan, Wyatt, & Hill, 2005; Thompson & Namey, 1990; Warburton, Nicol, & Bredin, 2006). Activity monitoring and biofeedback have been shown to increase physical activity (Bravata et al., 2007; Marcus et al., 1998) yet tools for monitoring wheelchair specific parameters of physical activity are limited.

Wheelchair users are also at risk for developing pressure ulcers from sitting too long without adequate pressure relief for tissue perfusion (Henzel, Bogie, Guihan, & Ho, 2011; Linder-Ganz, Yarnitzky, Yizhar, Siev-Ner, & Gefen, 2009). Individuals with spinal cord injury (SCI) are at a particularly high risk with an annual incidence of 31% (Garber, Rintala, Hart, & Fuhrer, 2000) and a lifetime incidence of 85% (Sumiya, Kawamura, Tokuhiko, & Ogata, 1997). There are currently 276,000 Americans living with SCI, and this population grows by roughly 12,500 annually ("UAB Spinal Cord Injury Model System," 2015). Pressure ulcers have been linked to increased morbidity and mortality (Krause & Saunders, 2011). Medical or surgical repair of established pressure ulcers is difficult and costly; up to \$70,000 for full thickness ulcers (Chan et al., 2013; Garber & Rintala, 2003; Garber et al., 2000; Stroupe et al., 2011).

To prevent the formation of pressure ulcers, wheelchair users need to intermittently relieve interface pressure by performing wheelchair depression raises and/or trunk leans ("Pressure Ulcer Prevention and Treatment Following Spinal Cord Injury," 2001). One approach to this problem is to provide a simple reminder (i.e., an alarm) when pressure relief exercises need to be completed. However, this approach is not sufficient because patients who attempt pressure relief maneuvers often do not achieve an adequate magnitude or duration of tissue unloading (Makhsous et al., 2007; Sprigle & Sonenblum, 2011). A system to measure pressure relief maneuvers in a home setting and provide reminders and feedback on frequency and success of this important preventative movement activity could reduce the incidence of pressure ulcers.

Existing technologies for pressure monitoring and pressure ulcer prevention are too costly and are not well adapted for use in tele-health (Brace & Schubart, 2010; Kim, Ho, Wang, & Bogie, 2010; Verbunt & Bartneck, 2010; Yang, Chou, Hsu, & Chang, 2010). The state of the art technology for monitoring wheelchair seat pressure is a high density pressure mat. This technology consists of thousands of pressure sensors on a seat pad. While useful for research and seating center clinic purposes, these pressure mapping systems are too costly for widespread use. In addition, they require a connection to a personal computer for data collection, and are not compatible with smartphones or tablets; as such, users cannot view

their data in real time and cannot receive biofeedback based on their current seat pressure from their mobile devices.

Previous approaches to wheelchair activity monitoring have been based on measuring angular kinematics or revolutions of the rear wheel to calculate speed and distance traveled (Coulter et al., 2011; Tolerico et al., 2007; Wilson, Hasler, Dall, & Granat, 2008). These devices cannot distinguish passive from active wheelchair movement. This problem has been addressed by assessing wheelchair propulsion based on kinematic measurements at the wrist and/or chest (S. Hiremath, Ding, Farrington, Vyas, & Cooper, 2013; S. V. Hiremath, Intille, Kelleher, Cooper, & Ding, 2015; Nooijen, de Groot, Stam, van den Berg-Emons, & Bussmann, 2015; Postma et al., 2005; Warms & Belza, 2004; Washburn & Caopay, 1999). However, this approach necessitates the use of wearable sensors.

The purpose of this study was to demonstrate the feasibility of a device for monitoring pressure relief maneuvers and physical activity for wheelchair users. The device quantifies physical activity by counting the number of wheel-pushes based on wheelchair acceleration. The device measures pressure relief maneuvers using a seat-shaped sensor (e.g., “seat-sensor”) consisting of three force sensing resistors (FSRs). A Bluetooth module communicates data to a smartphone app for charting. The system attaches to any wheelchair without modification.

Methods

Data Collection

Ten wheelchair users with spinal cord injury (38.6 ± 9.1 years old; 10 male; 0 female) and 10 non-disabled individuals (45.3 ± 10.8 years old; 7 male; 3 female) were recruited for this study. All participants were at least 18 years of age. Those subjects with SCI were full-time wheelchair users with community mobility (ASIA A-C), were between 2 and 20 years post SCI, and were able to perform pressure relief maneuvers without assistance. Subjects used their own seat cushions for the testing. The seat cushions used were made of either foam, gel or air, and were similar to the majority of commercially available wheelchair seat cushions. All subjects provided written informed consent prior to the experiment. All procedures were approved by the Rancho Los Amigos National Rehabilitation Center (RLANRC) Internal Review Board.

The wheelchair activity and pressure relief monitoring device consisted of a seat-sensor containing three FSRs (Sensitronics, WA, USA), an LIS3DH low-power tri-axial accelerometer (STMicroelectronics, Switzerland), a BLE112 Bluetooth module for communication (Bluegiga, TX, USA), and an in-house built smartphone app (Figure 1). The tri-axial accelerometer was housed in an enclosure that was rigidly attached to the frame of the wheelchair such that the positive directions of the three sensing axes were forward, right, and down. Acceleration was recorded at 50 Hz and FSR data at 10 Hz.

To demonstrate the feasibility of the seat-sensor for the detection of pressure relief maneuvers, all subjects completed three wheelchair depression raises, three forward trunk leans, and six lateral trunk leans (three towards each side). Figure 2 shows a subject

performing the four pressure relief maneuvers. The goal of these maneuvers was to relieve pressure under the ischial tuberosities in order to allow the tissue to reperfuse. Under normal usage, pressure should be relieved for 30 seconds every 30 minutes (“Pressure Ulcer Prevention and Treatment Following Spinal Cord Injury,” 2001). The subjects performed the 12 pressure relief maneuvers consecutively in a pre-defined order. The subjects received verbal instruction and visual demonstration as needed to perform the maneuvers. They were instructed to remove as much pressure as possible to achieve sufficient pressure relief during each maneuver, which was confirmed by a trained therapist. To avoid overtaxing the study participants, the duration of each maneuver was limited to 15 seconds. The subjects also rested for 5-10 seconds between maneuvers. During the maneuvers the seat-sensor was placed underneath the user's own seat cushion to simulate normal usage. An important design criterion of the device was that it could not interfere with the interface between the patient and their own seat cushion. A high density pressure mat (Vista Medical, Canada) was placed over the wheelchair cushion to measure the true pressure between the cushion and the subject. The high-density pressure mat sampled pressure data at 25 Hz.

To demonstrate the feasibility of wheel-push counting, the ten full-time wheelchair users navigated a flat 50 m outdoor track and a 100 m outdoor obstacle course under self-propulsion (e.g., wheel-pushes) and under assisted-propulsion (e.g., someone else pushing the wheelchair, no wheel-pushes). The 100m outdoor obstacle course contained concrete and grass terrain, inclined and declined ramps, and a curb. Each subject performed three self-propulsion trials and three assisted propulsion trials over the track and the obstacle course for a total of 60 trials. Subjects were asked to complete both courses at their normal speed and propulsion rate. During all trials the participants' wheelchairs were outfitted with the device and an instrumented push-rim to directly measure the torque applied to each wheel (SmartWheel, Out-Front, Mesa, AZ) (Asato, Cooper, Robertson, & Ster, 1993). The pushrim data were sampled at 100 Hz.

Detecting Pressure Relief Maneuvers

Analog data from the three FSRs in the seat-sensor (front, right, and left sensors; Figure 1) were normalized such that a value of zero represented no weight on the FSR sensor (open circuit), and a value of one represented maximum weight on the FSR sensor (closed circuit).

A subject-specific adaptive baseline for each FSR was determined by computing a moving average of FSR values (window of 200 seconds) during time periods when the subject was sitting stationary in the wheelchair (Figure 3). Stationary was defined as minimal variation of the left and right FSR sensors between the previous sample and the current sample. The adaptive nature of the baselines accounted for subjects shifting their body position in the wheelchair over time. The subject-specific nature of the baselines accounted for the varying densities of the subjects' seat cushions, as the seat-sensor was placed underneath the seat cushion so as not to modify the interface between the user and the seat.

To distinguish between the four pressure relief maneuvers, differences between the current FSR data and the baselines were calculated for each FSR at each time-point. These differences are referred to as *front*, *left*, and *right* for the front, left, and right FSRs, respectively. A depression raise was defined by *front*, *right*, and *left* all < -0.18 from

baseline (unitless difference between normalized FSR values). A forward lean was defined by *right* < -0.10, *left* < -0.10, and *front* > 0 from baseline. A right lean was defined by *left* < -0.08 and *right* > 0 (and vice versa for a left lean) from baseline. A pressure relief maneuver was considered to be successful if one of these conditions was met for at least 10 seconds. Pauses of less than 1 second in the detected pressure relief maneuvers were allowed, but longer pauses caused a pressure relief attempt to be considered unsuccessful. These thresholds were determined experimentally from the pilot test data to maximize pressure relief detection accuracy compared to the direct measure of interface pressure using a high-density pressure mat placed directly between the subject and the seat cushion. Using the pressure mat data, a pressure relief maneuver was considered successful if the average pressure recorded by the high-density pressure mat in a 7 in by 7 in region under each ischial tuberosity was less than or equal to 30 mmHg for 10 seconds or more (Bogie, Nuseibeh, & Bader, 1992; Rithalia & Gonsalkorale, 2000). This threshold was chosen because keeping the superficial tissue pressure of the ischial tuberosities below 32 mmHg is the traditional design criterion for a successful wheelchair seat cushion (Peterson & Adkins, 1982). According to Gefen, the pressure ulcer prevention industry uses interfacial pressure mapping to design wheelchair cushions and bed mattresses that produce contact pressure peaks lower than 32 mmHg (Gefen, 2007).

Detecting Wheel Pushes

Tri-axial acceleration signals were low-pass filtered at 3 Hz using a 7th order Butterworth filter. Filtered horizontal plane accelerations (i.e., the accelerometer axes pointing in the directions of forward and left lateral) were summed to create a singular acceleration vector; this sum is herein referred to as net horizontal plane acceleration (a_h). Based on the mounting orientation of the sensor, the left lateral acceleration had a minimal contribution to the net signal and could be excluded in future studies. Vertical acceleration was excluded as there is minimal vertical acceleration during wheelchair propulsion. Peaks in a_h were defined as maximum valued points between two zero crossings. To account for between subject differences in acceleration magnitude, a subject-specific threshold was defined as 40% of the average value of the peaks of a_h . Between subject differences in acceleration magnitude reflect differences in self-propulsion styles, body-weight, arm strength, and other factors. Peaks that were lower than the subject-specific threshold were excluded from further analysis.

Four descriptors (e.g., features) were computed for each of the remaining peaks. Three of these features were based on acceleration: the peak acceleration (g), the shortest time duration to a neighboring peak (seconds), and the peak secondary deceleration (g) defined as the lowest acceleration between the subsequent two zero crossings after the peak acceleration (Figure 4).

The fourth feature was based on variability in the normalized FSR data. First, the norm of the FSR signals from the sensors under the left and right ischial tuberosities was computed. Then, the fourth feature was defined as the average of the preceding five point-to-point differences in this norm, according to the following equations:

$$nFSR = \sqrt{lFSR^2 + rFSR^2} \quad \text{Equation 1}$$

$$Feat_4 = (\sum_{i=4}^i |nFSR_i - nFSR_{i-1}|) / 5 \quad \text{Equation 2}$$

where *lFSR* and *rFSR* represent the signals from the FSRs under the left and right ischial tuberosities, respectively.

For each feature vector (e.g., the four feature values associated with a particular peak in a_h), the probability of the feature vector being from a self-propulsion trial was computed using Bayes' theorem and assumptions of conditional independence as follows:

$$p(S|X) = p(S) \cdot \prod_{i=1}^4 p(x_i|S) \quad \text{Equation 3}$$

where $p(S|X)$ is the probability of the feature vector X being from a self-propulsion trial; $p(S)$ is the probability of any peak being from a self-propulsion trial (e.g., prior probability), which was estimated to be 0.5; and $p(x_i|S)$ is the probability of the i^{th} feature value given the observed distribution of i^{th} feature values for all a_h peaks observed during self-propulsion trials. The product of all the $p(x_i|S)$ terms is the joint probability. A probability threshold was then set such that peaks having a probability above this threshold were estimated to be the result of a wheel push and peaks having a probability below this threshold were estimated to be the result of wheelchair dynamics not caused by a wheel push, such as during assisted-propulsion or during coasting.

A 10-fold cross validation was used to test classifier performance. The set of feature vectors was randomly separated into 10 equal groups. For each group $p(x_i|S)$ values (Equation 3) were computed based on the observed distributions of feature values from the other 90% of data. 10-fold cross validation is a method to prevent over training and to ensure that a classifier is robust for future samples, not just for observed samples.

Results

Detecting Pressure Relief Maneuvers

Ten full time wheelchair users and ten non-disabled individuals attempted a total of 240 pressure relief maneuvers; three depression-raise attempts, three forward lean attempts, three left lean attempts, and three right lean attempts. Each attempt was labelled as successful or unsuccessful based on pressure recorded by the high-density pressure mat placed on the wheelchair seat cushion that provided a direct measure of interface pressure. A maneuver was successful if the pressure recorded by the high-density pressure mat under the ischial tuberosities was less than 30 mmHg (Bogie et al., 1992; Rithalia & Gonsalkorale, 2000).

Overall, 187 of 240 (77.9%) of pressure relief attempts were successful. The success rates by type were 100% for depression raises, 75.0% for left leans, 86.7% for right leans, and 50.0% for forward leans; 12 forward leans (20%) resulted in successful pressure relief under either the left or right ischial tuberosities but not both. Each successful maneuver was then categorized as depression raise, front lean, left lean, or right lean. For example, a forward lean attempt that resulted in successful pressure relief under the right ischial tuberosity but not under the left ischial tuberosity was labelled as a left lean.

Each exercise attempt was then analyzed and categorized as depression raise, front lean, left lean, or right lean using the data from the FSR seat-sensor that was placed under the user's seat cushion. These categories were compared to those based on the directly measured interface pressure using a confusion matrix. Of the 240 pressure relief maneuvers attempted by the wheelchair users and the healthy controls, 225 were properly categorized by the seat-sensor (accuracy: 0.94, sensitivity: 0.96, specificity: 0.80). Individually, all four types of maneuvers had accuracy above 0.94. Sensitivity was highest for depression raises (0.98) and lowest for front lean maneuvers (0.80). The category confusion matrix is shown in Table 1.

Detecting Wheel Pushes

During self-propulsion trials a total of 2,162 pushes were measured with the instrumented push-rim (893 during trials on the 50 m flat track and 1,269 during trials on the obstacle course). The developed wheel-push detection algorithm was compared to this ground truth measurement.

In total, 3,694 peaks in a_h were identified by the peak detection algorithm; 629 during assisted-propulsion on the flat 50 m track, 1032 during self-propulsion on the flat 50 m track, 626 during assisted-propulsion on the obstacle course, and 1407 during self-propulsion on the obstacle course. Four feature values were computed for each detected peak: the peak acceleration (g), the shortest duration to a neighboring peak (seconds), the peak deceleration (g), and a metric of variability in FSR data (Equations 1 and 2). Probability distributions for each feature are shown in Figure 5.

The 10-fold cross validation yielded 2112 pushes measured by the wheelchair activity monitor during self-propulsion trials (97.7%). This consisted of 906 pushes during trials on the 50 m flat track (101.5%), and 1206 pushes during trials on the obstacle course (95.3%). During assisted-propulsion trials there were 477 incorrectly identified pushes (8.0 per trial), of which 147 were from trials on the flat track and 330 were from trials on the obstacle course.

Discussion

This study demonstrates the validity of a novel device for monitoring pressure relief maneuvers and physical activity during self-propulsion and assisted-propulsion for wheelchair users. Our next steps are to assess the accuracy of the device during real-world use and then to couple the tested device with in-app reminders and charting in order to test whether this biofeedback can improve compliance with a prescribed pressure relief schedule

and/or increase daily physical activity. In addition, remote access to pressure relief data will be implemented, allowing caregivers to remotely monitor adherence.

In this study we sought to develop a very low computational cost classification algorithm that could be embedded into low-power sensor firmware by using a simplified version of naïve Bayesian classification to detect wheelchair propulsions. In a traditional naïve Bayesian classification, one would compute the joint probability of a feature vector given each class (e.g., assisted-propulsion or self-propulsion), and then compare these two values to predict the true class associated with the feature vector. For our simplified version, we computed only the joint probability of a feature vector assuming the class was self-propulsion, and then compared this value to a threshold to predict the true class. Naïve Bayesian Classification is only one of a number of low computational cost classification methods. In this study we selected Naïve Bayes because it was relatively easy to implement and it met our needs. Other classification methods were also tested; in particular, Support Vector Machine Classification performed very well. However, the number of support vectors could not be reduced sufficiently to make the Support Vector Machine approach feasible for embedded processing. Future studies could explore the relative efficacy of alternative approaches.

Furthermore, this study differed from a conventional classification study in that the ground truth was not known for each observation. That is, there were more identified peaks during self-propulsion trials than there were true wheel-pushes (as measured by the instrumented push rim). Peaks in wheel torque (measured with the push-rim) cannot be mapped one-to-one with peaks in acceleration. Therefore, we could not assign a ground truth value to each identified acceleration peak. Rather, our objective was to select a probability threshold that yielded an overall number of detected wheel pushes that most closely matched the true number of pushes measured using the instrumented push-rim during self-propulsion trials, while minimizing the number of wheel-pushes detected during assisted propulsion trials.

It is difficult to identify the incidence of pressure ulcers in community dwelling wheelchair users because self-reporting often leads to substantial underreporting (Phillips, Temkin, Vesmarovich, Burns, & Idleman, 1999). Pressure ulcers can be accurately diagnosed via telephone and/or video conference interactions between care providers and patients (Hill, Cronkite, Ota, Yao, & Kiratli, 2009). If conducted regularly these interactions have been shown to help with early diagnosis (Phillips et al., 1999). However, regularly interacting with every wheelchair user via telephone is not feasible. Detecting non-adherence with recommended pressure relief maneuvers would allow care providers to intervene prior to ulcer development, which would potentially have a significant positive impact on patient care and health care costs. Overall, tele-health technology is an increasingly prevalent tool for facilitating safe discharge into the community, preventing secondary complications, and encouraging patient-centered healthy lifestyles (Prvu Bettger & Stineman, 2007; Schwamm, Audebert, et al., 2009; Schwamm, Holloway, et al., 2009).

Patients with SCI comprise only a fraction of the population that could potentially benefit from a telehealth technology to monitor wheelchair seat pressure. More than 70% of nursing home residents spend at least part of their day in wheelchairs (Wick & Zanni, 2007). Many

of these individuals must rely on caregivers for positional changes to prevent pressure ulcer development. As a result, the prevalence of pressure ulcers in nursing homes has been estimated to range between 7 and 23% (Smith, 1995). A system based on our technology could provide feedback to health-care workers in nursing homes regarding the extent and timing of interface pressure for physically dependent persons. This system would be modified to accommodate the unique challenges of a nursing home environment, such as eliminating disruptive alarms and integrating wandering alerts.

There are several limitations of this study. We did not evaluate the performance of the telehealth monitor in real-world conditions. The wheel push counting algorithm was not evaluated during real-world scenarios such as indoor movement, during transfers into and out of the chair, and while riding in a vehicle. Indoor wheelchair propulsion is characterized by low velocity and low acceleration, which may not be detected by the algorithm. Vehicle travel may result in high acceleration that is erroneously detected as propulsion pushes. The pressure relief maneuver algorithm was not tested with stationary sitting lasting more than a few minutes, or during a full day of normal sitting. The subjects used their own wheelchair seat cushions; as such, variations in cushion material could impact the measurements recorded by the sensor mat. The same dataset was evaluated to empirically determine the thresholds and then determine the accuracy for both pressure relief detection and wheelchair propulsions. Future testing should be conducted to validate the threshold values in a new patient population and should focus on assessing the accuracy of the device during real-world use.

Conclusion

In closing, this research demonstrated the feasibility of monitoring pressure relief maneuvers and physical activity for wheelchair users using a simple technology. Such an approach, when coupled with telehealth systems, could have a broad and significant impact on the health of wheelchair users by helping to prevent pressure ulcers and a variety of other comorbidities associated with a lack of physical activity.

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Figure 1.

Components of the developed wheelchair activity and pressure relief monitoring device consisting of (top left) a seat-pad instrumented with three force sensing resistors, (bottom left) a sensing unit containing a tri-axial accelerometer and a Bluetooth module for communication, and (right) a smartphone app.

Figure 2.

An exemplary subject demonstrating the pressure relief maneuvers: (A) sitting still, (B) wheelchair depression raise, (C) forward lean, and (D) lateral lean. Each pressure relief maneuver was performed for 10 seconds.

Figure 3.

Normalized FSR signals (solid lines) and corresponding baseline signals (dashed lines) from an exemplary subject performing a wheelchair depression raise, left lean, right lean, and front lean. The black shapes are plotted at the end of the pressure relief maneuver.

Figure 4.

A four second sample of net horizontal plane acceleration (a_h) for an exemplary subject during a self-propulsion trial on the 50m track. For the second peak (green circle) the three features values would be the following: the magnitude of the positive peak (green circle, 0.24 g); the magnitude of the negative peak (red circle, -0.25 g) between the subsequent two zero crossings (blue circles); and the shortest duration to a neighboring peak (e.g., the smaller of d1 or d2).

Figure 5. Probability density functions for feature values during self-propulsion (solid green) and assisted-propulsion (dashed red).

Confusion matrix for classification of pressure relief attempts. Rows correspond to actual classes based on interface pressure measured using a high-density pressure mat. Columns correspond to predicted classes based on the seat-sensor containing three FSRs placed under the subject's self-selected seat cushion.

Table 1

		Seat-Sensor (placed under seat cushion)					
		Raise	Left Lean	Right Lean	Front Lean	No Exercise	Total
Interface Pressure (ground truth)	Raise	59	1	0	0	0	60
	Left Lean	0	39	0	2	5	46
	Right Lean	0	1	59	2	1	63
	Front Lean	0	1	4	24	1	30
	No Exercise	0	0	4	4	33	41
	Total	59	42	67	32	40	240