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A wearable system for pre-impact fall detection

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ABSTRACT

Unique features of body segment kinematics in falls and activities of daily living (ADL) are applied to make automatic detection of a fall in its descending phase, prior to impact, possible. Fall-related injuries can thus be prevented or reduced by deploying fall impact reduction systems, such as an inflatable airbag for hip protection, before the impact. In this application, the authors propose the following hypothesis: “Thigh segments normally do not exceed a certain threshold angle to the side and forward directions in ADL, whereas this abnormal behavior occurs during a fall activity”. Torso and thigh wearable inertial sensors (3D accelerometer and 2D gyroscope) are used and the whole system is based on a body area network (BAN) for the comfort of the wearer during a long term application. The hypothesis was validated in an experiment with 21 young healthy volunteers performing both normal ADL and fall activities. Results show that falls could be detected with an average lead-time of 700 ms before the impact occurs, with no false alarms (100% specificity), a sensitivity of 95.2%. This is the longest lead-time achieved so far in pre-impact fall detection.

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1. Introduction

Falls are a major care and cost burden to health and social services world-wide (Annekenny and O’Shea, 2002). Falls have traditionally been recognized as one of the “giants” for geriatric medicine, reflected from the high incidence of falls, common adverse sequelae such as fractures, and the major psychological impact. Among the causes of falls, fainting (syncope) is one common factor in older people and also related to unexplained and recurrent falls (McIntosh et al., 1993). Syncopal episodes or fainting related falls are unwitnessed in 40–60% of older people over 65 (McIntosh et al., 1993) and cause considerable mortality and morbidity among this age group.

Even though most falls produce no serious injury, only 1–2% of falls result in hip fractures (Hayes et al., 1996), 5–10% of community-dwelling older adults who fall each year do sustain a serious injury such as fracture, head injury, or serious laceration (Nevitt et al., 1991; Tinetti et al., 1995). Of all the fall-related traumas, fractures of the neck and trochanteric regions of the femur, the major bone in the hip joint, are currently one of the most serious health care problems faced in aging populations (Marks et al., 2003). Most hip fractures (60–99%) are related to direct trauma to the hip (Chapuy et al., 1992; Cummings and

Nevitt, 1989; Hipp et al., 1991; Lauritzen and Askegaard, 1992). An investigation performed by Smeesters et al. showed that at any gait speed, that faint falls resulted in a greater number of sideways falls with impact near the hip (Smeesters et al., 2001). In these scenarios, the most promising prevention strategies for faint fall involves the identification of individuals who are at increased risk and the implementation of appropriate interventions, these include physical restraint (Gross et al., 1990), investigation of fall-related fractures prevention strategies (Smeesters et al., 2001; Van den Kroonenberg et al., 1996; Yamamoto et al., 2006), study of characteristics and risk factors of syncope (Kenny and O’Shea, 2002; Peczalski et al., 2006), and multi-factorial risk assessment and management (Sjösten et al., 2007; Weatherall, 2004).

In fall intervention strategies, one of the key concerns in preventing or reducing the severity of injury in the elderly is to detect the fall in its descending phase (Hayes et al., 1996) before the impact (pre-impact fall detection). A few groups have attempted to detect falls prior to impact (Bourke et al., 2008; Nyan et al., 2006; Wu, 2000). Some researchers have investigated inflatable hip protectors to cushion the fall prior to impact (Davidson, 2004; Lockhart, 2006; Ulert, 2002). Wu implemented pre-impact fall detection by thresholding the horizontal and vertical velocity profiles of the trunk using motion analysis system. Wu showed that falls can be distinguished from activities of daily living (ADL) with 300–400 ms lead-time before the impact (Wu, 2000). Nyan et al. used three gyroscope sensors at three different locations, the sternum, front of the waist and under the

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arm, for fall pre-impact detection. Nyan achieved 100% sensitivity with approximately 200 ms lead-time before the impact; however, 16% of ADL events tested were misinterpreted as falls (Nyan et al., 2006). In addition, at the instant when fall is detected, the angle of body configuration from the vertical axis is 40–54° (Nyan et al., 2006). Bourke et al. investigated pre-impact detection of falls by thresholding the vertical velocity of the trunk. An optical motion capture system and an inertial sensor unit consisting of a tri-axial accelerometer and a tri-axial gyroscope were used in their experiments. The inertial sensor was located on the chest of the body using a harness. Falls can be distinguished from normal ADL, with 100% accuracy and with an average of 323 ms prior to trunk impact and 140 ms prior to knee impact, in that subject group (Bourke et al., 2008).

In pre-impact fall detection, if a fall can be detected in its earliest stage, i.e., in the descent phase (Hayes et al., 1996), more efficient impact reduction systems can be implemented with a longer lead-time for injury minimization.

This paper presents the implementation and clinical trial results of a wearable pre-impact fall detection prototype, using inertial sensors to detect faint falls in its incipience. The approach is based on the characteristics of angular movements of the thigh and torso segments in falls and ADL. The authors hypothesize the following statement: “Thigh segments normally do not exceed a certain threshold angle to the side and forward directions in ADL, whereas this abnormal behavior occurs during a fall activity”. This pre-impact fall detection algorithm can be implemented in a wearable fall injury minimization system to track a user’s body movement and notify the fall impact reduction device when to activate in order to reduce the severity of the fall injury.

2. Materials and methods

The hardware setup developed for the pre-impact fall detection prototype includes a thigh sensor set (TS), waist sensor set (WS) and data processing unit (Fig. 1a). The TS contains one Freescale¹ MMA7260Q (± 4 g, 300 mV/g) tri-axial micromachined accelerometer (x : vertical (downward positive); y : lateral (right positive); z : sagittal (forward positive)) and two Analog Devices² ADXRS150 ($\pm 150^\circ/s$) rate gyroscopes measuring pitch (back positive) and roll (left positive) angular velocity. The sensitive axes of the sensors are shown in the figure (Fig. 1b). A single tri-axial accelerometer is included in the WS with similar sensitivity axis setting as those in TS. Two AAA-size batteries can power the sensor unit for nearly two days. Acceleration data ($x_{TS}(i)$, $y_{TS}(i)$, $z_{TS}(i)$, $x_{WS}(i)$, $y_{WS}(i)$, $z_{WS}(i)$) are low pass filtered with cutoff frequency of 0.5 Hz and the gyroscope signals ($\omega_{LAT}(i)$, $\omega_{SAG}(i)$) are band-pass filtered between 0.5 and 2.5 Hz. Hardware RC low-pass and band-pass filters are used in implementation to avoid time delay and phase shift in digital filtering. The sampled data are then smoothed with a simple exponentially weighted moving average filter.

Chipcon CC2420 Zigbee transceivers are used for data communication between the sensor sets and data processing unit. An Intel[®] PXA255 processor (400 MHz) is used in the data processing unit. Three AAA-size batteries can power the processing unit for 12 h. Sensor data is sampled at sampling rate of 47 samples/s. In wireless communication between the transmitters and the receiver, polling medium access control (MAC) was used. The polling MAC is designed mainly for the multi-transmitter BAN systems requiring the continuous transfer of data at a central processing point. The receiver at the processing unit sends a polling packet to each transmitter at the two sensor sets in a round robin fashion. Upon reception of the polling packet, the transmitter will be allowed to transfer a pending packet immediately. There are two light emitting diodes (LEDs), green and red, to show the status of the system. The red LED is used to show the data communication status between the sensor sets and the processing unit and the green LED turns on for 30 s when a fall is detected. For post-trial analysis, data was also recorded onto a laptop through a Zigbee transceiver, at the laptop’s serial port, during the experiment.

The process flow of pre-impact detection algorithm in the processing unit is shown in Fig. 2. Acceleration samples are transformed into two dimensional degrees of body orientation, measuring how many degrees these body segments deviate from the vertical axis (i.e., standing is 0° and lying flat on the floor is 90°),

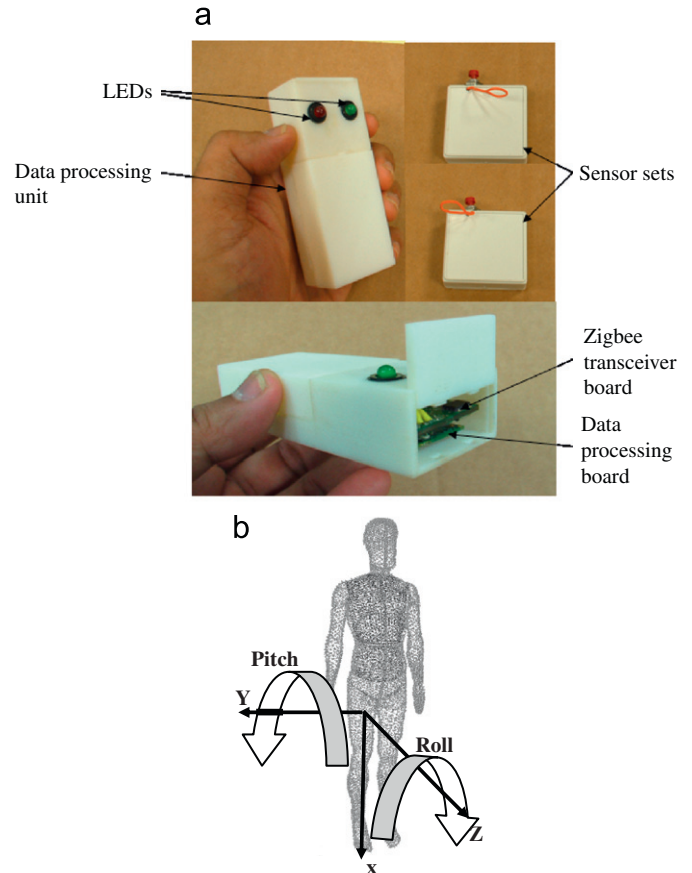


Fig. 1. (a) Hardware setup of the pre-impact fall detection system. Data processing board ($8 \times 3.5 \times 2$ cm, 70 g), sensor board ($4.3 \times 4.3 \times 1.5$ cm). (b) Sensitivity axes of the 3D accelerometer and 2D gyroscope. The TS is attached using Velcro at the front of the right thigh and the WS is attached on a belt at the front waist position.

using the following equations:

$$\text{deg}_{SAG,TSorWS}(\text{backward positive}) = -\tan^{-1}(z_{TSorWS}/x_{TSorWS}) * (180/\pi), \quad (2.1)$$

and

$$\text{deg}_{LAT,TSorWS}(\text{left positive}) = \tan^{-1}(y_{TSorWS}/\sqrt{1-y_{TSorWS}^2}) * (180/\pi), \quad (2.2)$$

where TS and WS represent thigh segment and torso segment. The 3-D accelerometer’s degree conversion is calibrated with 30–60–90 and 45–45–90 triangle metal blocks located on a small table whose surface is adjusted to be horizontal with a carpenter’s spirit level. For calibration of the ADXRS150 rate gyroscope sensor, 12.5 mV/°/s was used to convert gyroscopes’ outputs (raw voltages) to degree/second measurement, this value was obtained from the manufacturer’s data sheet. Angular data (deg_{SAG} , deg_{LAT}) were examined according to the hypothesis, if one of two dimensional angular signals ($\text{deg}_{SAG,TS}$ and $\text{deg}_{LAT,TS}$) of the thigh segment intersects the threshold levels, $\pm 10^\circ$, then a fall is confirmed using the following algorithm. First, the correlation coefficient (ρ_{deg}) of thigh angular data (TAD) and torso angular data (WAD) are compared (i.e., between $\{\text{deg}_{SAG,TS}(-N+i); i = 0, \dots, N \text{ and } N = 19\}$ and $\{\text{deg}_{SAG,WS}(-N+i); i = 0, \dots, N \text{ and } N = 19\}$, or between $\text{deg}_{LAT,TS}$ and $\text{deg}_{LAT,WS}$) to the threshold value ($\rho_{deg} \geq 0.8$). Secondly, if the correlation coefficient (ρ_{ω}) between the band-pass filtered gyroscope segment $\{\omega(-N+i); i = 0, \dots, N \text{ and } N = 79\}$, and its corresponding reference template is greater than or equal to 0.8, then a fall is confirmed. The reference templates are those taken from experimental data used in the development stage of the algorithm before the experiment whose results are presented in this paper was conducted. The experiment conducted during the development stage is described in the Appendix A. Thus, falls were simulated similarly to those presented in the paper to generate the reference templates. If $\text{deg}_{LAT,TS}$ intersects the $+10^\circ/-10^\circ$ threshold level, the gyroscope segment is correlated with left/right fall reference template and if $\text{deg}_{SAG,TS}$ intersects the $+10^\circ/-10^\circ$ threshold level, the gyroscope segment is correlated with backward/forward fall reference template. The LED will switch on if the coefficients ρ_{deg} and ρ_{ω} are above or equal to 0.8 (Fig. 2). In this fall detection algorithm, only the thigh angular data between 0° to $\pm 90^\circ$ are taken as points of interest for the

¹ Freescale Semiconductor, Inc., Austin, TX, USA.

² Analog Devices, Inc., Norwood, MA, USA.

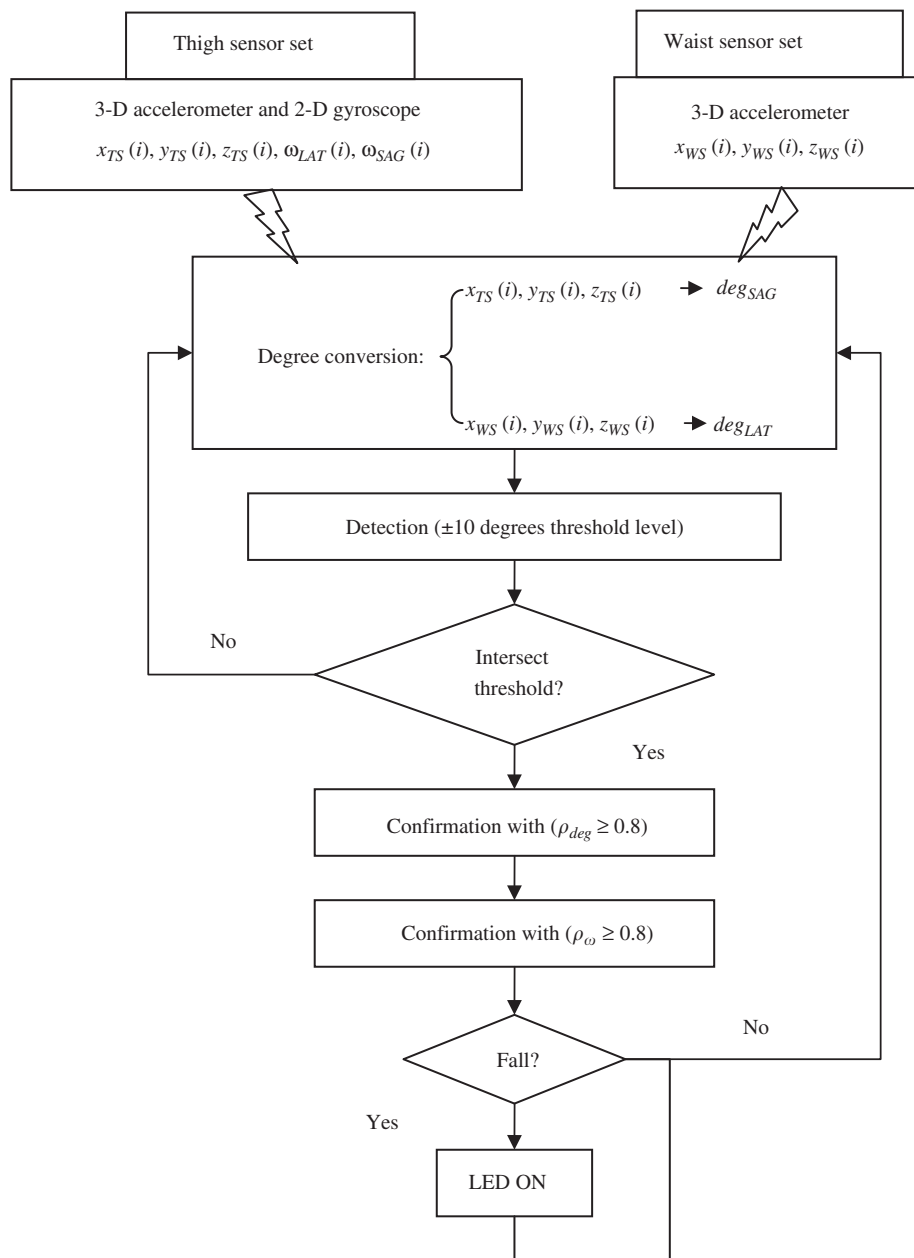


Fig. 2. Fall pre-impact detection algorithm.

confirmation process. Here, the term “0 to ± 90 ” refers only to the trend of thigh segment’s sway in fall activities and it does not mean that thigh rotates only between 0 and ± 90 in fall activities. The whole algorithm, written in the C programming language, can be processed in less than 7 ms using Intel® PXA255 (400 MHz) processor.

Thirteen male and eight female volunteers participated in the clinical trial. The average ages were 23.38 and 22.25 years old, respectively, with a range of 22–27 for males and 21–for females. The mean height and mass \pm standard deviation of the males were 1.76 ± 0.06 m and 69.1 ± 8.72 kg, respectively. For the female volunteers, the mean height \pm standard deviation was 1.65 ± 0.03 m, and for mass it was 51.58 ± 4.47 kg. The trial protocol was approved by the Singapore General Hospital Medical Research Ethics Committee and written informed consent was obtained from each subject. In faint fall simulations, the subjects were told to stand on the floor beside the mattress and simply relax themselves and fall to the sides, back, and front. The subjects performed the simulated fainting incidents on a 6 in thick soft foam mattress. For ADL, a chair, the mattress and two flights of stairs were used for sitting, sit–stand transitions, walking, stand–sit transitions, lying, ascending and descending stairs. Each activity was conducted twice and recorded using a camcorder (Panasonic, VDR-D300GC) with a frame rate of 30 frames/s. The observer simultaneously reset the camcorder and the processing unit during the experiment.

3. Results

Angular movements of thigh and torso segments, their respective correlation coefficient data (moving window size is 20 samples), and gyroscope data for lateral and sagittal movements of ADL conducted by a subject in the experiment are shown in Figs. 3 and 4. ADL were conducted using the following procedures:

- (1) Segment a: Initially the subject was sitting down on a chair (referring to thigh sagittal angular data (TSAD) (Fig. 3a) which is at approximately 90° and the torso sagittal angular data (WSAD) (Fig. 3b) is at 0°). The subject then stood up and walked. A sit–stand transition activity can be seen at TSAD (Fig. 3a), which shows transition between approximately 90° and 0° and a few similar cycles of gyroscope data (Fig. 3f) represents a walking activity.

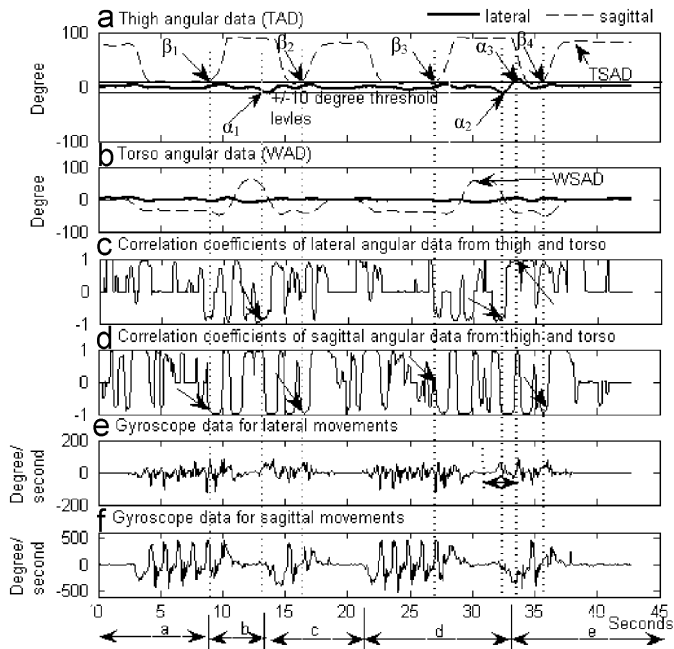


Fig. 3. Angular data, correlation coefficients (moving window size is 20 samples) and gyroscope data for normal activities.

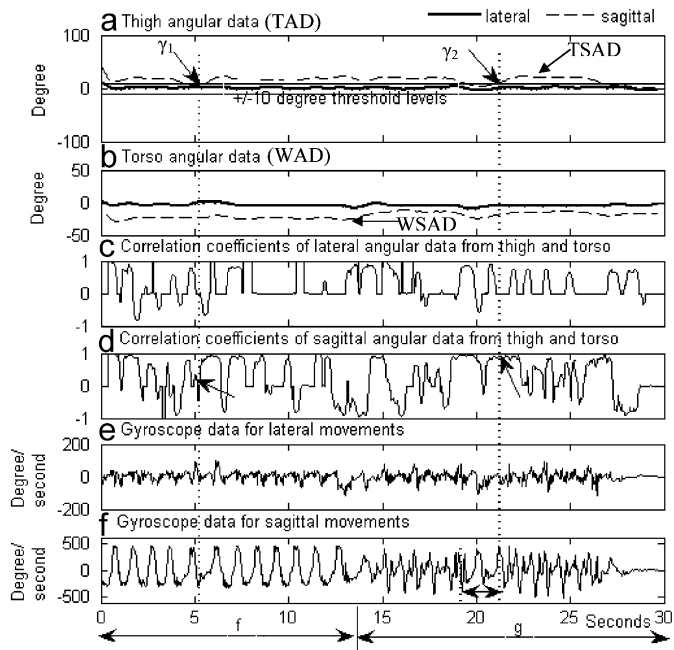


Fig. 4. Angular data, correlation coefficients (moving window size is 20 samples) and gyroscope data for normal activities.

- (2) Segment b: The subject then sat down on the mattress (referring to the transition between 0° and 90° at TSAD (Fig. 3a) and lay down (referring to TSAD of Fig. 3a and WSAD of Fig. 3b which are around 90°).
- (3) Segment c: Next activities were, standing up (referring to transition between 90° and 0° at TSAD (Fig. 3a), walking for two or three steps (as discussed in segment a) and sitting down on the chair (as discussed in segment a).
- (4) Segments d and e: The subject repeated the activities as shown in segments a–c in the same sequence.

- (5) Segments f and g: After that the subject ascended the stairs for 12 steps and descended the stairs. Repetitive cycles of gyroscope's sagittal data represent stairs activities (Fig. 4f).

Activities were validated with recorded video clips in the experiment. According to the hypothesis “thigh segments normally do not exceed a certain threshold angle to the side and forward directions in ADL”, only three intersection points (α_1 , α_2 , and α_3) were found between TSAD (lateral movements) and threshold levels and no intersection points were found between TSAD and -10° threshold level (forward movements) (Figs. 3a and 4a). The number of intersection points can be reduced by increasing the threshold level, but it is kept as low as possible with the compromise between lower false alarm frequency and a longer lead-time interval, i.e., to get longer lead-times, the threshold level is kept as low as possible. As thigh rotates from 0° to $\pm 90^\circ$ in fall activities, only the intersection points where thigh angular data go from 0° to $\pm 90^\circ$ are the point of interests in this study. Altogether nine intersection points (α_{1-3} , β_{1-4} , $\gamma_{1,2}$) were found, as pointed by arrows in Figs. 3a and 4a; most of them were caused by backward movements (β_1 , β_2 , β_3 , and β_4) in sitting down activities. However, almost all of them could be rejected as false alarms since their respective correlation coefficients were lower than the threshold value ($\rho_{deg} = 0.8$) as arrowed in Figs. 3c, d and 4c, d. Conversely the second last and last arrows in Figs. 3c and 4d (related to arrows α_3 and γ_2) were two possibilities for false alarms, but they were rejected again by ρ_ω as they were -0.6 (Fig. 3e) and 0.332 (Fig. 4f), i.e., lower than the threshold level ($\rho_\omega = 0.8$). The ρ_ω values were computed between gyroscope segments, shown by double sided arrows (Figs. 3e and 4f), and their respective reference templates (Fig. 5). In the left side fall activity (Fig. 6), the person started to fall down at about 9 s and ρ_{deg} was well above the threshold level. Then, ρ_ω between the segment shown by double sided arrow (Fig. 6d) and the left side fall reference template was 0.997. Correlation coefficient relationships from falls and normal activities were plotted (Fig. 7). A total of 42 data points were plotted for falls and 216 data points were plotted for normal activities. Two hundred and sixteen is the number of detected points using $\pm 10^\circ$ threshold levels for all falls and ADL performed by the 21 subjects. No false alarms were found in the experiment (100% specificity). All falls, except for two front falls, could be detected; thus, a sensitivity of 95.2% (40/42) was

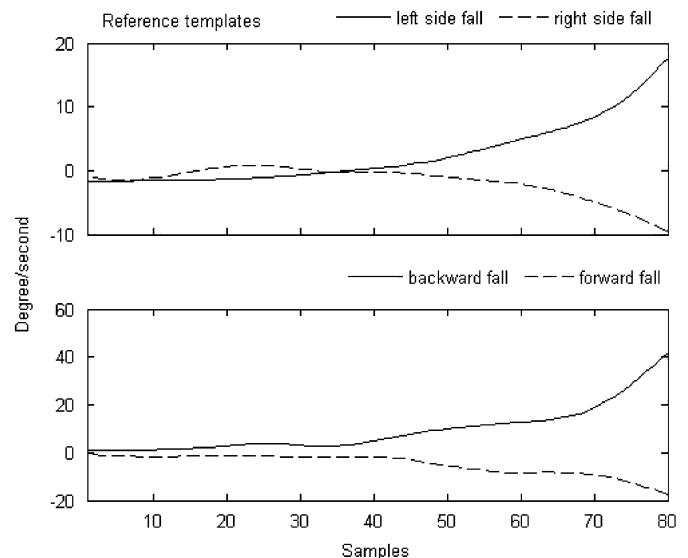


Fig. 5. Fall reference templates used in ρ_ω computation.

achieved. Means and standard deviations of lead times for the different fall types are shown in Fig. 8.

4. Discussion

As most of the fall-related injuries occur when the body hits the ground (Wu, 2000), the application of a pre-impact fall

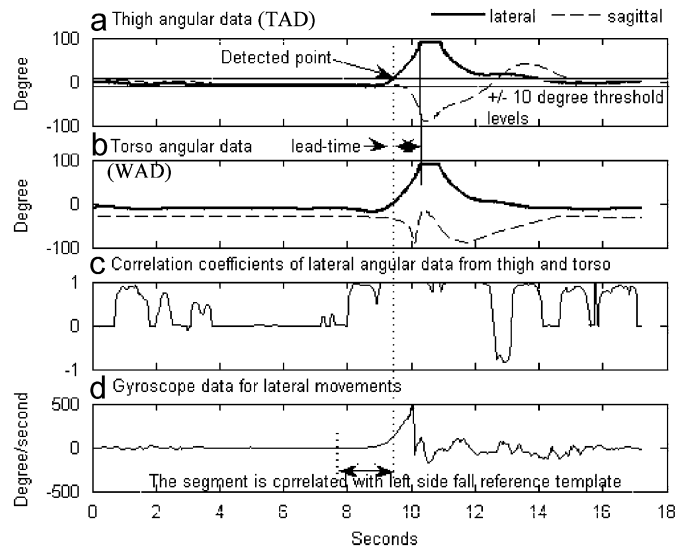


Fig. 6. Angular data, correlation coefficients (moving window size is 20 samples) and gyroscope data for fall activity.

detection approach along with fall impact reduction systems (Davidson, 2004; Lockhart, 2006; Ulert, 2002) for injury minimization, will provide useful intervention for elderly people susceptible to faint falls.

This study aimed to detect a fall in its inception for a longer lead-time using the unique feature of body segments encountered in falls and ADL. In this study, we achieved a lead-time of 700 ms, this is the longest lead-time obtained so far in pre-impact fall detection (Wu, 2000; Bourke et al., 2008; Nyan et al., 2006).

Detection of thigh segment's movement in the forward and sideways directions is very unique in pre-impact fall detection. Most of the intersection points between the thigh angular data and threshold levels ($\pm 10^\circ$) in the ADL performed were caused by

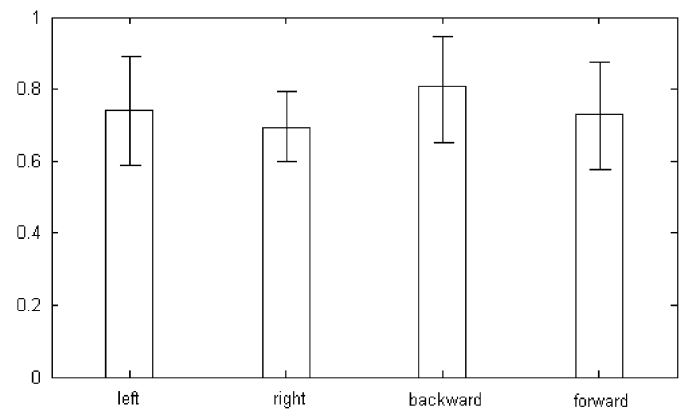


Fig. 8. Means and standard deviations of lead times for fall activities.

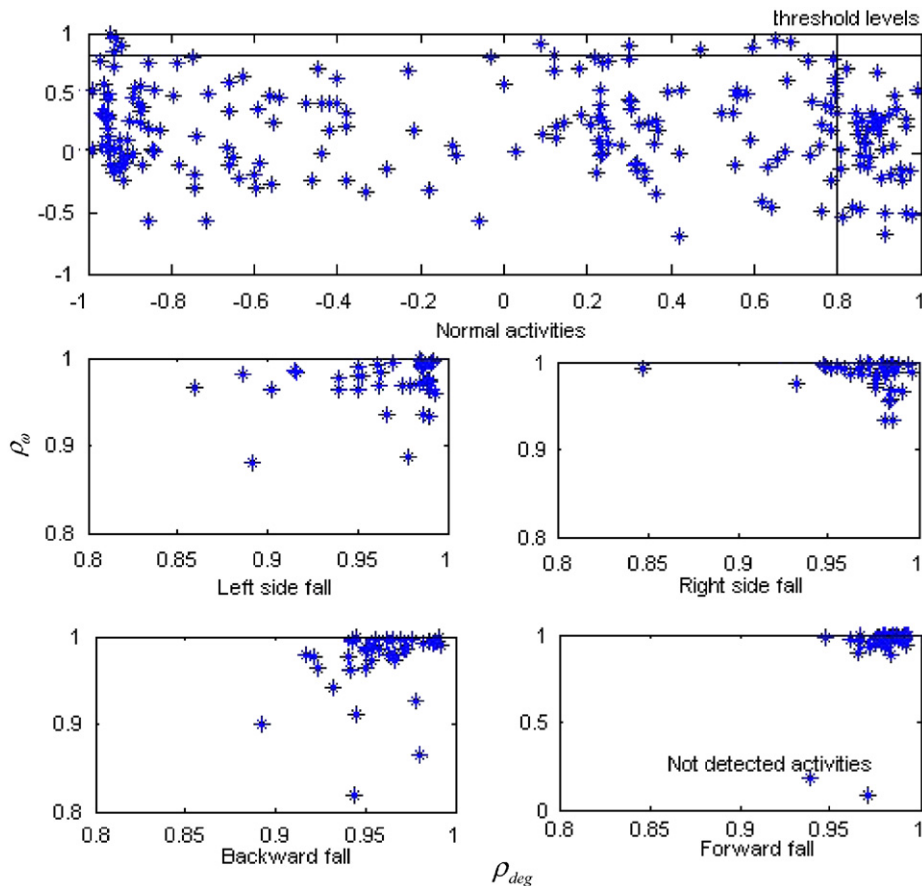


Fig. 7. ρ_{deg} and ρ_w relationships of falls and normal activities.

backward movements such as the sitting down activity. By increasing the threshold level, the total number of intersection points, 216, can be reduced. However threshold levels of $\pm 10^\circ$ were retained as it produced the optimum level with the compromise between lead-time and the number of false positives. For backward movements such as the sitting down activity, the possibility of false alarm can be rejected using the correlation coefficient of the angular data from both segments as they move in opposite directions during this activity. Moreover, as presented in the materials and methods section, only the thigh angular data swaying from 0° to $\pm 90^\circ$ are taken as points of interest for the confirmation process. Here, “ 0° to $\pm 90^\circ$ ” refers only the trend of thigh segment’s sway in fall activities. It does not mean that the thigh rotates only between 0° and $\pm 90^\circ$ in fall activities. As $\pm 10^\circ$ is the threshold in fall detection, a fall is considered only for the angular data trend of the movement from 0° to $\pm 90^\circ$, i.e., the data trend representing the movement from $\pm 90^\circ$ to 0° such as data found in sit–stand transition is not considered in detection. That is also a way of false alarm prevention in fall pre-impact detection.

There have been active research attempts to study fall detection using acceleration and velocity characteristics, but none so far exploit the correlation of the movement of body segments. Using correlation results in a more robust, longer lead-time and accurate methodology when daily activities are taken into consideration. In detection, a movement can be considered as a fall if ρ_{deg} and ρ_{ω} are above the threshold of 0.8. The algorithm was tested in a clinical trial with day-to-day activities that elderly people normally perform. As a result, all fall activities were detected apart from two forward falls; however, no false positives occurred. The reason for the misdetection of these two activities is due to the fact that the two subjects first knelt down before they released their bodies onto the mattress in the forward direction. Before they released their bodies forward, the movement of thigh segments was more similar to a sitting down activity than a forward fall. An average lead-time of 700 ms before the impact from a fall was achieved. This is the longest lead-time achieved so far in pre-impact fall detection (Bourke et al., 2008; Nyan et al., 2006; Wu, 2000).

In the assessment of successful balance recovery from complete loss of balance in fall (Madigan and Lloyd, 2005; Thelen et al., 1997; Wojcik et al., 1999), Thelen et al. found that the maximum lean angle where subjects could recover balance with a single forward step averaged 32.5° for young men and 23.9° for older men (Thelen et al., 1997). In his experiment, a horizontal lean-control cable was attached to the back of a padded pelvic belt to support the subjects while they kept their bodies approximately straight in a forward-leaning posture before they were released for forward falls. Madigan and Lloyd found similar results for their male subjects (29.9° and 20.5° , respectively) (Madigan and Lloyd, 2005). Wojcik et al. found that the maximum lean angle averaged 30.7° for young women and 16.2° for older women (Wojcik et al., 1999). Therefore, it can be noted that our threshold level of $\pm 10^\circ$ of thigh angle is well within the limits of balance recovery ability in the fall process.

It should be pointed out that all activities tested in this experiment were performed by healthy volunteers aged below 30, as the experimental procedure is understandably not suited for elderly subjects who are at greater risk of suffering injury. The movement of younger subjects is bound to differ from that of the elderly population, who may have a slower reaction time and lesser ability to rescue the body from falling. However, thigh segment’s movement detection and correlation of body segments’ movements are less affected by this limitation as it is a postural condition independent of physique. Moreover, the proposed hypothesis and the developed algorithm were tested against a small range of fall types, i.e., faint fall, among those that elderly

people encounter in their daily lives. Therefore, further tests are needed for other types of falls such as falls preceded by walking such as tripping and slipping. An appropriate temperature compensation strategy is also required as temperature drift may affect the performance of the system during a long term implementation.

In conclusion, a pre-impact fall detection system has been developed that will allow detection of an impending fall, as a result of a faint, with a lead-time of 700 ms before impact occurs to the vulnerable areas of the body. This is the longest lead-time achieved so far in pre-impact fall detection. Subsequently through the implementation of effective fall impact reduction systems, such as an inflatable airbag for hip protection, it is envisaged that injuries that occur to the elderly as a result of a fall can be reduced or even prevented.

Conflict of interest statement

I declare that I have no proprietary, financial, professional, or other personal interest of any nature or kind in any product, service and/or company that could be construed as influencing the position presented in, or the review of, the manuscript entitled, “Wearable system for pre-impact fall detection”.

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Appendix A. Supplementary data

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jbiomech.2008.08.009

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