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Determination of simple thresholds for accelerometry-based parameters for fall detection

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***Abstract*—The increasing population of elderly people is mainly living in a home-dwelling environment and needs applications to support their independency and safety. Falls are one of the major health risks that affect the quality of life among older adults. Body attached accelerometers have been used to detect falls. The placement of the accelerometric sensor as well as the fall detection algorithms are still under investigation. The aim of the present pilot study was to determine acceleration thresholds for fall detection, using triaxial accelerometric measurements at the waist, wrist, and head. Intentional falls (forward, backward, and lateral) and activities of daily living (ADL) were performed by two voluntary subjects. The results showed that measurements from the waist and head have potential to distinguish between falls and ADL. Especially, when the simple threshold-based detection was combined with posture detection after the fall, the sensitivity and specificity of fall detection were up to 100 %. On the contrary, the wrist did not appear to be an optimal site for fall detection.**

I. INTRODUCTION

THE increasing population of elderly people (aged 65+) is mainly living in a home-dwelling environment and needs applications to support their independency and safety. Falls are one of the major health risks that affect the quality of life among elderly by producing fear and resulting in decrease in mobility and activity [1]-[3]. It appears that elderly people are willing to accept new technologies to support their independence and safety [4], [5].

Body attached accelerometers [6]-[9] and gyroscopes [10], [11] have been used to detect human movement and especially falls. The placement of an acceleration sensor to optimize the location of a fall detector has been studied in some extent. The placement site at the waist has been suggested to be the most efficient, since at this site the acceleration signal is similar and evenly distributed between different fall types [1]. Furthermore, waist attached accelerometers are located near to the body center of gravity providing reliable information on subject's movements, with the exception of movements of arms and legs [12].

The sum vector of a triaxial accelerometer signal has been suggested to be more accurate in fall detection than single axis thresholds [10]. Start of the fall, before the actual impact, has been monitored using the norm of the triaxial signal [13], [14] or velocity [13], [15]. The actual impact has been detected with threshold based algorithms [6], [14], [16].

The aim of the present pilot study was to determine acceleration thresholds for fall detection, using triaxial accelerometric measurements at the waist, wrist, and head.

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II. MATERIALS AND METHODS

A. Subjects, Test Falls and ADL

Intentional falls and activities of daily living (ADL) were performed by two voluntary subjects (aged 22 and 38 years). Forward, backward, and lateral falls were performed towards an air-filled bed or a combination of a tatami and mattresses. Acceleration signal was measured either from the waist (falls n=14, ADL n=31), wrist (falls n=12, ADL n=15), or head (falls n=5, ADL n=12). ADL samples represented dynamic activities (e.g. walking, walking on the stairs, picking up object from the floor) and posture transitions.

B. Accelerometry

Accelerations during the falls and ADL were measured with body attached triaxial accelerometers, constructed using three uniaxial capacitive accelerometers [17]. Each triaxial accelerometer was attached to a separate data logger (Onset Computer Corp. Tattletale 8v2) with the sampling frequency of 400 Hz [17].

Accelerometers were attached to the non-dominant wrist, same side of the waist, and in front of the forehead. The sensitive axes of the head and waist worn accelerometers were mediolateral, anteroposterior, and vertical. The axes of the wrist worn accelerometer were identical when the subject stressed his/her arms (palm downwards) to the side.

C. Data Processing

Accelerometer data were loaded from data loggers to a computer (software CrossCut 2.01, Borland International), and converted into gravitational units with a custom-made

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MATLAB program. Data processing, analyses and fall detection simulation were done with a custom-made LabVIEW (8.0) program. Each of the three devices was calibrated to generate acceleration signals of 1g and -1g when parallel to the positive and negative axis of the gravitation, respectively.

The measured acceleration signal was processed by resampling at 50 Hz and median filtering (window length 3) to reduce the data amount and noise before any further analyses. The processed data were low-pass (LP) or high-pass (HP) filtered (f_c= 0.25 Hz) with a digital second order Butterworth filter when necessary. The LP data were used in posture analyses and HP data in motion analyses.

D. Parameters

Total sum vector SV_{TOT} (Fig. 1a), containing both the dynamic and static acceleration components, was calculated from resampled data as indicated in (1).

$$SV = \sqrt{(A_x)^2 + (A_y)^2 + (A_z)^2}, \quad (1)$$

where A_x, A_y, A_z = acceleration (g) in x-, y-, and z-axes. The start of the fall was determined as the pit before the impact, SV_{TOT} being equal or lower than 0.6 g. Dynamic sum vector SV_D was calculated similarly from the HP filtered data by using (1).

Fast changes in the acceleration signal were investigated by constructing a new sum vector SV_{maxmin} (Fig. 1b), which was calculated using the differences between the maximum and minimum acceleration values in a 0.6 s sliding window for each axis. Vertical acceleration Z_2 was calculated as indicated in (2).

$$Z_2 = \frac{SV_{TOT}^{\ 2} - SV_D^{\ 2} - G^2}{2G}, \quad (2)$$

where SV_{TOT} = total sum vector (g), SV_D= dynamic sum-vector (g), and G = gravitational component = 1 g.

The time period between the start of a fall and the impact was determined from the measurements at the waist, by recognizing the minimum value of SV_{TOT} in the pit ($SV_{TOT} <$ 0.6 g) and the impact-related maximum peak of SV_{TOT} or Z_2.

Velocity v_0, just before the fall associated impact, was calculated by integrating the area around the pit (Fig. 1A) where the SV_{TOT} was lower than 1 g. The posture was detected 2 seconds after the impact using the LP filtered vertical signal [6], [18].

The threshold values for different parameters were adjusted to optimal detection of falls with minimized false alarms from ADL samples (maximal sensitivity with 100% specificity when possible). For fall detection, lying posture after fall was required. The posture determination was not used with data measured from the wrist.

For comparison with previous studies, falling index (FI) was also calculated for the waist measurements as described earlier by Yoshida et al. [19] using a time window of 0.4 s.

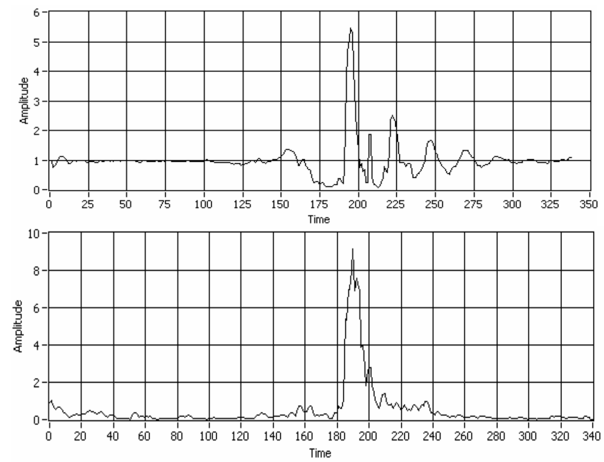

Fig. 1. (a) SV_{TOT}. The start of the fall can be seen as the pit before the impact peak, and impact is detected as the highest peak. (b) SV_{maxmin}. The impact is detected as the highest peak.

III. Results

A. Waist

When measured from the waist, the value ranges of the parameters SV_{TOT}, SV_D, SV_{maxmin}, and Z_2 were slightly overlapping between falls and ADL (mean and quartile values shown in Fig. 2). The threshold values set for waist worn application are shown in Fig. 2 and summarized in Table I. When posture detection after the fall was included, the specificity of fall detection was 100% for all parameters (Table I). The velocity v_0 before the impact ranged from 0.8 to 3.4 ms^{-1} with an average value of 1.7 ms^{-1}, 95% of the falls having velocities over 1.0 ms^{-1}. On average, the impact was detected 0.3 s after the beginning of the fall, but the time range was up to 1 s. Falling index FI had a value range from 2.65 to 5.89 g and from 1.10 to 3.62 g during falls and ADL, respectively, with an overall fall detection sensitivity of 59%.

B. Wrist

The value range of parameters SV_{TOT}, SV_{maxmin}, and SV_D from wrist worn device overlapped clearly between falls and ADL (Fig. 3). Z_2 values overlapped slightly, but here 75% of falls had higher value than the maximum from ADL (Fig. 3). Threshold values are shown in Fig. 3 and summarized in Table I. The velocity v_0 before the impact varied between 0.2 and 2.9 ms^{-1} with an average value 1.0 ms^{-1}.

C. Head

From the head, the value ranges of SV_{TOT}, SV_{maxmin}, SV_D, and Z_2 had specific value ranges for falls and ADL with no overlapping (data not shown). Threshold values are summarized in Table I. The specificity of fall detection was 100% for all parameters (Table I). The velocity v_0 before the impact ranged from 0.8 to 2.3 ms^{-1} with an average value of 1.7 ms^{-1}.

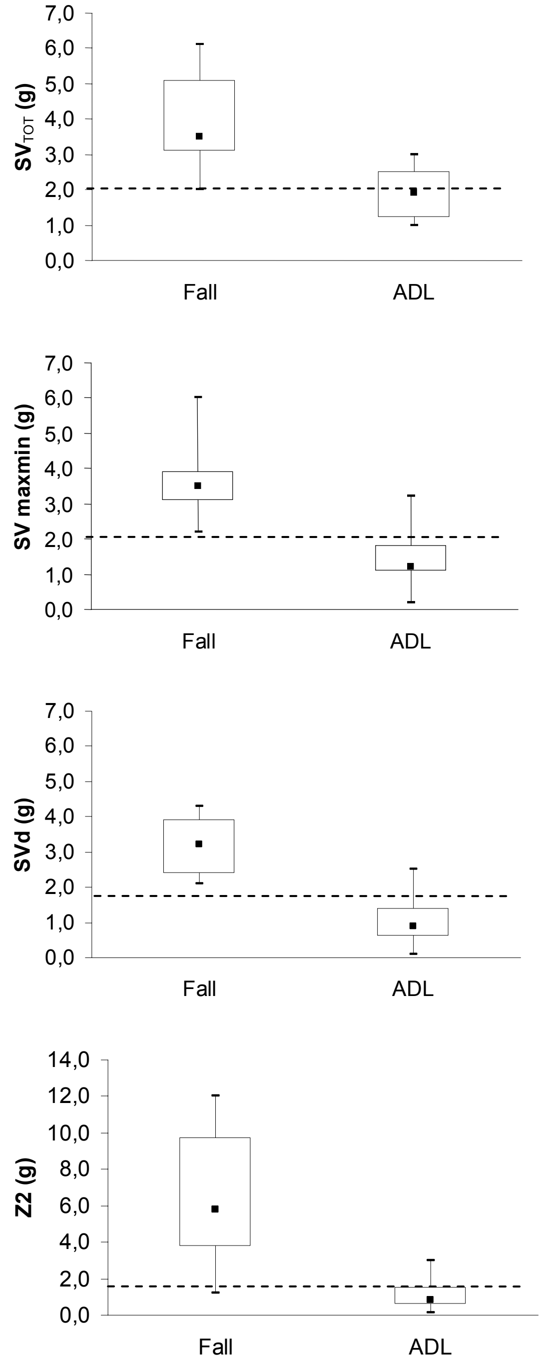

Fig. 2. Quartile box plots of parameters measured from waist during falls and ADL. Selected threshold values (th) are marked (---). SV_{TOT} (th 2.0 g), SV_{maxmin} (th 2.0 g), SV_D (th 1.7 g), and Z_2 (th 1.5 g).

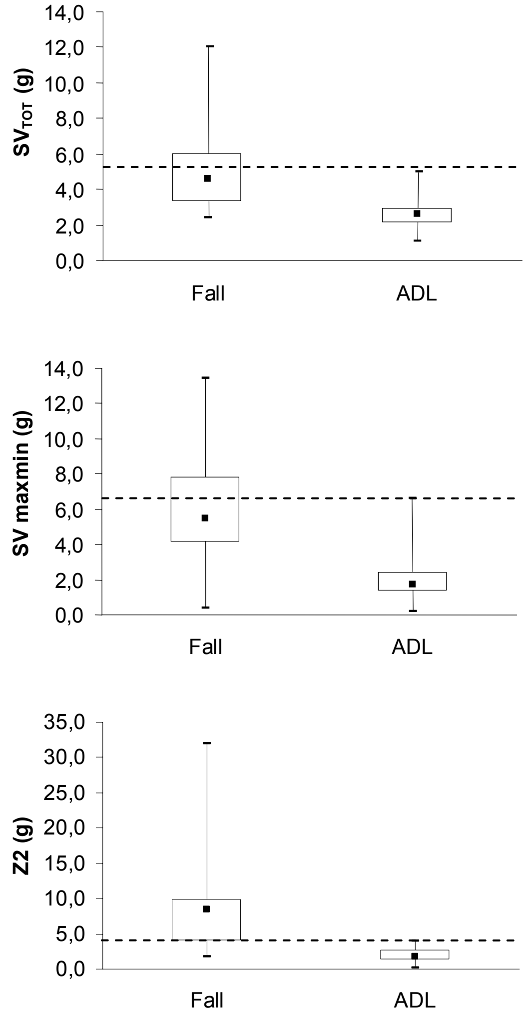

Fig. 3. Quartile box plots of parameters measured from wrist during falls and ADL. Selected threshold values (th) are marked (---). SV_{TOT} (th 5.2 g), SV_{maxmin} (th 6.5 g), and Z_2 (th 3.9 g).

IV. Discussion

This study investigated the acceleration signal measured with body attached accelerometers from intentional falls and activities of daily living (ADL) to determine threshold values for multiple parameters capable of discriminating between falls and ADL.

Our results showed that even if the different parameters measured from the waist showed typical characteristics for fall and ADL, the value ranges had some overlapping. This indicates that using simple thresholds alone is not optimal for practical fall detection. This is contrary to the report of Bourke et al. [14] where they were able to determine a simple SV_{TOT} threshold value capable of discriminating between falls and ADL with 100% sensitivity and specificity. However, their experimental procedure used young test subjects for fall events and elderly for ADL, whereas we used same subjects for both samples. When we included the posture detection after the fall, the thresholds obtained resulted in a sensitivity and specificity of 95-100%.

Our threshold for the acceleration sum vector (SV_{TOT}) from waist was 2.0 g, which was smaller than the threshold of 3.52 g presented earlier [14]. This difference might be partly explained by the median filtering used here, which changes the absolute peak value of the impact signal.

All tested parameters measured from the head were able to totally distinguish falls and ADL. This indicates that head-worn accelerometer would be a reasonable choice for fall detection, using e.g. hearing-aid housing as suggested

TABLE I
THRESHOLD VALUES FOR FALL DETECTION ALGORITHMS

	Waist		Wrist		Head	
Parameter	Th	Se / Sp (%)[a]	Th	Se / Sp (%)	Th	Se / Sp (%)
SV_{TOT}	2.0 g	100/100	5.2 g	45/100	2.0 g	100/100
SV_D	1.7 g	100/100	5.1 g	32/100	1.2 g	100/100
SV_{maxmin}	2.0 g	100/100	6.5 g	41/100	1.7 g	100/100
Z_2	1.5 g	95/100	3.9 g	75/100	1.8 g	100/100

Th = threshold, Se / Sp = sensitivity and specificity (%), g = acceleration of gravity. [a] Posture detection after fall included

[15], if there were no limitations in usability and acceptance.

It would also be possible to integrate a fall detector to a wrist watch [13]. Even though the usability of a wrist-worn fall detector is excellent, the acceleration signal measured from wrist varies widely. Here, the signal of ADL samples was strongly overlapping with that of falls. In addition, posture detection is not applicable in wrist-worn accelerometers. Thus, it appears that wrist is not an optimal site for fall detection, and this placement site would need the most complicated algorithm for fall detection as previously suggested [1]. However, the parameter Z_2 shows the highest potential to distinguish falls and ADL at the wrist.

At the waist, most of the falls had velocities at the typical range shown previously for intentional falls. These velocities are capable of resulting hip fractures among elderly [20]. At the head level, the minimum value of velocity was at the range that has been used as a threshold in fall detection algorithm by Lindemann et al. [15].

Testing our data with the earlier published FI threshold resulted in fall detection sensitivity being in good accordance with the sensitivity range of 40 to 100% reported by Yoshida et al. [19], supporting our test procedure.

This was a pilot study with only two test subjects. Thus, the results presented are only suggestive and experimental studies with a larger population, preferably including mid-aged or elderly subjects, are needed to confirm the results.

We conclude that head and waist are relevant sites for accelerometric detection of falls, using simple thresholds and posture detection. On the contrary, the wrist does not appear to be an optimal site for fall detection.

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