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An instrumented sit-to-stand test used to examine differences between older fallers and non-fallers

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***Abstract* — An instrumented version of the five-times-sit-to-stand test was performed in the homes of a group of older adults, categorised as fallers or non-fallers. Tri-axial accelerometers were secured to the sternum and anterior thigh of each participant during the assessment. Accelerometer data were then used to examine the timing of the movement, as well as the root mean squared amplitude, jerk and spectral edge frequency of the mediolateral (ML) acceleration during the total assessment, each sit-stand-sit component and each postural transition (sit-stand and stand-sit). Differences between fallers and non-fallers were examined for each parameter. Six parameters significantly discriminated between fallers and non-fallers: sit-stand time, ML acceleration for the total assessment, and the ML spectral edge frequency for the complete assessment, individual sit-stand-sit components, as well as sit-stand and stand-sit transitions. These results suggest that each of these derived parameters would provide improved discrimination of fallers from non-fallers, for the cohort examined, than the standard clinical measure – the total time to complete the assessment. These results indicate that accelerometry may enhance the utility of the five-times-sit-to-stand test when assessing falls risk.**

I. Introduction

The five-times-sit-to-stand test (FTSS) was first described by Csuka and McCarty in 1985 as a standardised measure of lower extremity strength [1]. It has since been established that it is strongly associated with postural balance disorders [2, 3], and that it is an independent predictor of falls risk [4, 5]. During the test, participants must stand up from a chair, and sit down again, five times as quickly as possible. The standard outcome measure is the stopwatch-measured time taken to complete the task.

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Recent studies have examined the use of FTSS to identify balance disorders and those at higher risk of falling. Whitney *et al.* [3] reported that discriminant analysis of the FTSS time could identify 65% of participants with a balance dysfunction from those without. Buatois *et al.* [4] reported that participants who took longer than 15 seconds to complete the FTSS had a 74% greater risk of recurrent falls than those who took less time. The same group reported that the FTSS provided more added value to a falls risk assessment than the "Timed up and go" test, and the "One-Leg-Balance" test, particularly when examining those at moderate risk of falls [6].

Using accelerometry it is possible to examine the amplitude and power spectrum of the acceleration of the body during the assessment in addition to temporal parameters. It also facilitates examination of the different phases of the test – the ability to stand from a chair is an important task for independent daily living, and the ability to sit into a chair from standing in a controlled fashion is equally important. Analysis of the time and frequency components of these movements for fallers and non-fallers may provide insight into the underlying cause of a participant's falls history.

Instrumenting the FTSS using body-worn sensors facilitates a more detailed analysis of the movement, with potential for improved classification of participants. Accelerometry has been shown to be a valid tool for examining a single sit-to-stand movement in studies comparing it to a video-based system [7-9] and a force platform [10]. Najafi *et al.* [5] used gyroscopes to detect and assess the duration of sit-stand and stand-sit transitions, and reported that the transition durations were significantly correlated with falls risk. A method to examine the postural transitions, sit-stand and stand-sit, during the timed up and go test has also been reported [11].

Narayanan *et al.* [12] examined the FTSS using accelerometers to develop a falls risk estimation tool, deriving a range of parameters related to the timing and acceleration of the trunk during the assessment. In that study, the dissimilarity between sit-stand-sit components was identified as a useful marker for falls, using a linear least squares model. Investigation of individual postural transitions, and the amplitude and frequency content of the acceleration during the corresponding time periods may identify further markers of falls risk.

The aim of this study was to examine differences between fallers and non-fallers using accelerometer-derived parameters of the FTSS, to identify which components of the assessment may have utility in identifying older adults at risk of falls.

II. Methods

This study was performed as part of a larger research project which examined diurnal variations in the outcome of different clinical assessments and their association with falls history, these results are in preparation for publication elsewhere. In this study, the body-worn sensor data captured during the FTSS assessments were examined, to quantitatively examine parameters which may relate to falls risk.

A. Participants

40 community-dwelling older adults (19 fallers (7 male), 20 non-fallers (9 male); age: 71.4 ±7.3 years; BMI: 27.19 ± 3.57 kg/m^2) gave their informed consent and participated in this study. Participants were recruited from the TRIL Clinic in St James's Hospital, Dublin. Ethical approval was obtained from the St. James's Hospital Ethics Committee.

Participants were categorised as fallers if they had fallen more than once in the previous five years, or if they had one fall which required medical attention, or if they suffered from fear of falling or one of the following cardiovascular risk factors: orthostatic hypotension, carotid sinus hypersensitivity or vasovagal syncope. Participants were categorised as non-fallers if they did not meet these criteria.

B. Protocol

The FTSS was conducted in the home under supervision four times during a single day. Participants were asked to refrain from vigorous exercise on the previous day and on the day of the experiment.

All participants were advised to eat a light breakfast between 0830-0900hr, light lunch at 1230hr and light snack at 1430hr, and were asked to refrain from consuming caffeinated drinks. They were advised to take their medications at the usual time and the timings of medications were recorded.

The FTSS was performed using two tri-axial accelerometers (Shimmer Research, Dublin, Ireland), one attached to the anterior of the right thigh, and one to the sternum. One accelerometer was positioned along the line joining the anterior spina iliaca superior to the superior part of the patella, such that one axis measured the superioinferior (SI) acceleration of the thigh. The other accelerometer was positioned above the sternum, such that one of its axis measured the mediolateral (ML) acceleration of the sternum during the assessments. A 46 cm high chair was used, and participants were asked to keep their arms folded across their chest for the duration of the assessment, as illustrated in Figure 1. Each participant was then asked to fully stand up and sit back down five times as quickly as possible.

Figure 1. Illustration of experimental set-up of the FTSS, with the participant in the initial position and sensors worn on sternum and right thigh.

C. Data acquisition

All data were sampled at 102.4 Hz and streamed via Bluetooth to a PC using a custom application developed in BioMOBIUS™, and were subsequently analysed using Matlab 7.10 (The Mathworks Inc., Natick, MA, USA).

D. Data analysis

Accelerometer data were calibrated using a standard procedure [13] and then band-pass filtered 0.1 – 5 Hz [14] using an 8th order Butterworth filter with a corner frequency of 50 Hz. The method described by Moe-Nilssen [15] was used to correct for the effects of gravity on the accelerometer data recorded at the sternum. This was not applied to the data recorded at the thigh due to the rotation of the recording axis during the assessment. A sample of processed accelerometer data from each sensor location is presented in Figure 2.

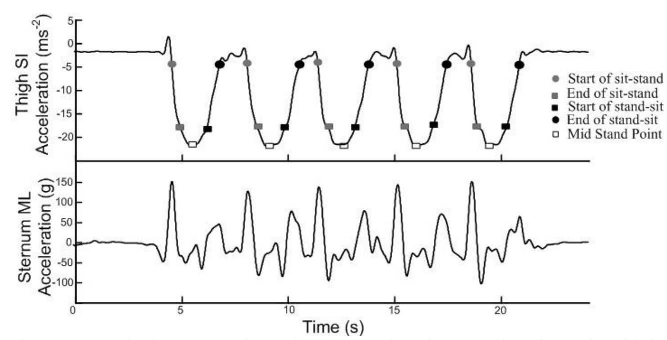

Figure 2. Typical processed accelerometer signals recording from the thigh (top) and from the sternum (bottom) during the FTSS test. Mid-stand points, and the start and end of each sit-stand and stand-sit phase are indicated using the superioinferior signal recorded at the thigh.

The superioinferior acceleration of the thigh was used to identify the key time points of the assessment. Each mid-stand point was first identified as the minimum acceleration for each sit-stand-sit trough, A_{MS}, Figure 2. The start and end of each standing phase, sitting phase and sit-stand-sit component were established using the empirically tuned thresholds $0.2A_{MS}$ and $0.8A_{MS}$ which ensured all five movements were detected, while unsuccessful attempts were not included in the postural transition duration.

When the signal amplitude decreased past $0.2A_{MS}$, this was taken to indicate the start of a standing phase. When the signal amplitude fell below $0.8A_{MS}$, this was taken to indicate the end of a standing phase. Similarly, when the signal amplitude increased above $0.8A_{MS}$, this indicated the start of a sitting phase, and when it increased above $0.2A_{MS}$, this indicated the end of a sitting phase, Figure 2.

The total time to complete the FTSS was calculated as the difference between the time at the end of the fifth sitting phase and the time at the start of the first standing phase. In addition, the mean and coefficient of variation (CV) of the time taken to complete individual sit-stand-sit components, sit-stand and stand-sit phases of each session were calculated.

The accelerometer signal recorded at the sternum was used to examine ML movement during the assessment. The active portion of the ML accelerometer signal was identified, defined as the section of the signal commencing at the start of the first sit-stand phase and terminating at end of the fifth stand-sit phase. The root mean squared (RMS) amplitude of this portion was used as a measure of total ML sway (g). The RMS amplitude of the ML signal during each sit-stand-sit component, sit-stand phase and stand-sit phase were also examined. The ML jerk of each portion of the test was also examined, calculated as the derivative of the acceleration with respect to time (g/s).

The steadiness of the mediolateral movement was examined using the spectral edge frequency, SEF, of the mediolateral accelerometer data. The SEF was defined as the frequency below which 95% of the power of the signal is contained, and was used here to provide an insight into the frequency content of the acceleration signal.

E. *Statistical analysis*

One-way analysis of variance (ANOVA) was used to examine whether each derived parameter varied significantly across the four repetitions for each participant. One-way ANOVA was then used to examine whether each derived parameter varied significantly between fallers and non-fallers. P-values less than 0.05 were considered statistically significant. The statistical power of all significant observations was calculated using GPower 3.1 [16].

III. RESULTS

The time, mediolateral RMS amplitude, mediolateral jerk and mediolateral spectral edge frequency were examined for the total test, individual sit-stand-sit components, sit-stand transitions and stand-sit transitions. Parameters did not vary significantly between repetitions (p>0.1).

Results for each parameter are presented for fallers and non-fallers in Table I, and boxplots for parameters which significantly discriminated between groups are presented in Figure 3.

Fallers took significantly longer to complete sit-stand transitions than non-fallers, Figure 3A. Fallers also exhibited increased jerk over the complete assessment than non-fallers, Figure 3B. Spectral edge frequency significantly discriminated fallers from non-fallers for the total test, sit-stand-sit components, sit-stand and stand-sit transitions. In all instances, fallers exhibited a higher spectral edge frequency than non-fallers, Figure 3C-F.

The statistical power of each significant observation was greater than 83%.

TABLE I
Accelerometer-derived parameters for each segment of the FTSS, presented for fallers and non-fallers. ANOVA results (F- and p-values) are presented for the difference between fallers and non-fallers

Parameter	**Non-fallers**	**Fallers**	**ANOVA**	
			F	**p**
Total time (s)	15.96±4.1	17.18±4.7	1.99	0.16
Mean sit-stand-sit time (s)	2.21±0.64	2.34±0.68	0.96	0.33
CV sit-stand-sit time (%)	11.13±10.06	10.9±9.23	0.01	0.91
Mean stand-sit time (s)	0.55±0.19	0.54±0.19	0.11	0.74
CV stand-sit time (%)	22.33±19.89	19.64±21.62	0.43	0.51
Mean sit-stand time (s)	0.41±0.2	0.49±0.18	4.25	**0.04**
CV sit-stand time (%)	26.89±29.01	26.39±26.27	0.01	0.93
Total RMS (g)	80.36±29.37	70.38±32.51	2.66	0.11
Mean sit-stand-sit RMS(g)	66.56±16.96	63.9±18.95	0.56	0.46
Mean stand-sit RMS (g)	75.85±17.45	77.46±18.62	0.20	0.65
Mean sit-stand RMS (g)	80.47±20.74	78.48±33.17	0.14	0.71
Total jerk (g/s)	-0.03±0.04	-0.01±0.03	5.13	**0.03**
Mean sit-stand-sit jerk (g/s)	-0.19±0.12	-0.16±0.18	1.70	0.20
Mean stand-sit jerk(g/s)	0.42±0.5	0.4±0.38	0.09	0.77
Mean sit-stand jerk(g/s)	-1.64±0.67	-1.53±0.91	0.52	0.47
Total SEF (Hz)	1.77±0.21	1.87±0.24	5.39	**0.02**
Mean sit-stand-sit SEF (Hz)	3.2±1.04	3.73±1.16	6.09	**0.02**
Mean stand-sit SEF (Hz)	13.3±3.81	15.13±3.52	6.23	**0.01**
Mean sit-stand SEF (Hz)	11.29±3.11	13.12±4.83	5.42	**0.02**

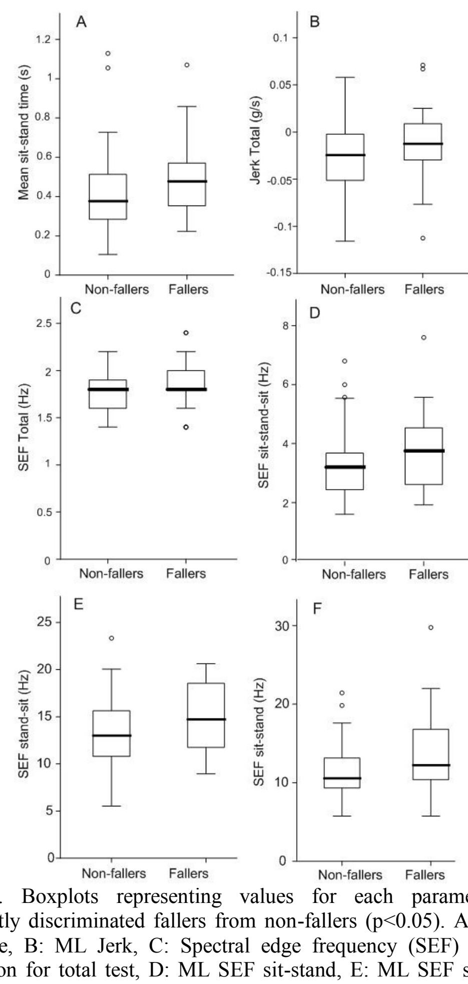

Figure 3. Boxplots representing values for each parameter which significantly discriminated fallers from non-fallers (p<0.05). A: Mean sit-stand time, B: ML Jerk, C: Spectral edge frequency (SEF) of the ML acceleration for total test, D: ML SEF sit-stand, E: ML SEF stand-sit; F: ML SEF sit-stand-sit components.

IV. Discussion

The time taken to complete the five-times-sit-to-stand test provides an indicator of falls risk and balance impairment [3, 6]. Body-worn accelerometers provide an inexpensive, easy to implement method of quantifying additional measures of timing and postural movement during the FTSS. This study examined a range of sensor-derived parameters associated with the FTSS, and components of the assessment, in order to examine differences between participants with a history of falls and those without.

Instrumentation of the FTSS using body-worn sensors may improve the accuracy of the assessment, while removing the human error associated with stopwatch measurements. However, limitations of body-worn accelerometers must also be considered, such as the sensitivity to sensor placement, the influence of gravity and potential difficulties with data collection. These issues may be overcome with due care in sensor placement and the by the use of reliable software.

Previous studies have demonstrated the value of the time taken to complete the FTSS in identifying those at risk of falls [3, 4, 6], with a time greater than 15 seconds reported by Buatois *et al.* to indicate increased falls risk [4]. The fallers examined here took longer to complete the FTSS than the non-fallers examined, however this measure did not significantly discriminate fallers from non-fallers. This study examined 40 participants in the home environment, which may not be sufficiently large to replicate the observations of Buatois *et al.* who examined 2735 participants in a clinical environment [4].

The results of this study indicate that fallers take significantly longer to complete sit-stand transitions compared to non-fallers, indicating reduced quadriceps and core muscle strength in this category. Najafi *et al.* [5] reported increased mean postural transition duration for fallers. However, they reported no significant difference between sit-stand and stand-sit durations, which may differ from the results presented here due to the low number of participants (eleven) in that study. The fallers examined in this study took longer to stand up, but no difference was observed in the time taken to sit down, Table I. Similar results have been observed for Parkinson's disease patients, when the sit-stand and stand-sit transitions during the "timed up and go" assessment were examined [11].

The temporal derivative of acceleration, or the jerk, over the total assessment was higher in fallers, indicating less controlled movement in this category. Additionally, the spectral edge frequency was significantly higher for fallers for the total assessment, sit-stand-sit components as well as stand-sit and sit-stand transitions. The higher frequency components suggest that fallers exhibit less smooth mediolateral sway, and that this variable may provide a useful insight when attempting to identify those at risk of falling.

The accelerometer-derived parameters which significantly discriminated between participant groups provided enhanced discrimination between the fallers and non-fallers examined in this study, compared with the accelerometer-derived surrogate of the standard clinical measure – the total time taken to complete the assessment. These results suggest that the parameters described in this study may enhance the utility of the FTSS, particularly when assessed in the home environment. Further research, potentially involving a larger dataset and additional analysis, would be required to extend this conclusion beyond the present cohort. Statistical models, accounting for gender and age, developed using the described parameters may further enhance the instrumented FTSS. The results presented here indicate that an instrumented FTSS may provide a suitable method for in-home, potentially unsupervised, monitoring of falls risk or balance assessment.

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