Development and validation of a physical activity monitor for use on a wheelchair

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Running Title: Wheelchair Physical Activity Monitoring

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Abstract

Purpose: Keeping physically active is important for people who mobilise using a wheelchair. However, current tools to measure physical activity in the wheelchair are either not validated or are limited in their application. The purpose of this study was to develop and validate a monitoring system to measure wheelchair movement. **Methods:** The system developed consisted of a tri-axial accelerometer placed on the wheel and an analysis algorithm to interpret the acceleration signals. The two accelerometer outputs in the plane of the wheel were used to calculate the angle of the wheel. From this, outcome measures of wheel revolutions, absolute angle, and duration of movement were derived and the direction of movement (forwards or backwards) could be distinguished. Concurrent validity was assessed in comparison with video analysis in 14 people with spinal cord injury using their wheelchair on an indoor track and outdoor wheelchair skills course. Validity was assessed using Intraclass-Correlation Coefficients (ICC(2,1)) and Bland Altman plots. **Results:** The monitoring system demonstrated excellent validity for wheel revolutions, absolute angle, and duration of movement (ICC(2,1) >0.999, 0.999, 0.981, respectively) for forwards and backwards direction, in both manual and powered wheelchairs, when the wheelchair was propelled forwards and backwards, and for movements of various durations. **Conclusion:** This study has found this monitoring system to be an accurate and objective tool for measuring detailed information on wheelchair movement and manoeuvring regardless of the propulsion technique, direction and speed.

Key Words:

Accelerometer, , validation studies, rehabilitation, exercise, movement, spinal cord injuries

Introduction

Physical inactivity has been identified as a major risk factor for a number of health complaints such as coronary heart disease, diabetes mellitus, osteoporosis, and some cancers[1]. Research on the effects of physical activity in the disabled population is limited. Although it is believed that people with disabilities can obtain health benefits from a physically active lifestyle, many do not participate in the recommended amount of health promoting exercise and physical activity[2, 3]. Wheelchair users have limited opportunity to be physically active and propelling a manual wheelchair is a primary means of physical activity and exercise.

Self report questionnaires have been widely used to measure physical activity[4, 5], however, it is well recognised that these are subject to difficulties with memory recall and overestimation[2, 6]. Different approaches to objectively measuring wheelchair locomotion, either position devices at the wrist (measuring wheelchair propulsion)[7, 8, 9], or on the wheelchair wheel (measuring wheelchair movement)[10, 11]. When placed at the wrists, accelerometer based monitors use algorithms to identify repetitive or bilateral wrist movement as wheelchair propulsion, separating it from other upper limb tasks[7]. Outcome measures include duration, activity counts and energy expenditure during manual propulsion of the wheelchair[7, 8, 9]. When placed on the wheelchair rear wheel, objective monitors may measure the distance, speed and duration of movement[10, 11]; and may be used with both powered and manual wheelchairs[11]. Current monitors are poor at assessing components of manoeuvring that are either small or do not consist of repetitive wrist movements. Using a monitor which measures all bouts of wheelchair movement in manual and powered wheelchairs could provide an insight into a person's rehabilitation and mobility levels in the home and community environments. This could be used to provide information on community accessibility and reintegration[10].

The purpose of this study was to develop and validate a technique for continuously monitoring wheelchair movement in both manual and powered wheelchair users. This technique consisting of a tri-axial accelerometer and analysis algorithm should be capable of providing detailed information of wheelchair mobility, clinically relevant outcome measures, and have the potential to be used for a wide application.

Methods

Instrument

The activPAL trio physical activity monitor (PAL Technologies Ltd, Glasgow, UK) is a single unit (5 x 3.5 x 0.7 cm, 20g), which consists of a tri-axial accelerometer, power source, real time clock, and data storage, and has a sampling frequency of 10Hz. This monitor can measure activity for up to 10 days and provides the same outcomes as the uni-axial activPAL physical activity monitor (PAL Technologies Ltd, Glasgow, UK), which was previously used in the wheelchair population[11], and is a valid and reliable instrument for measuring continuous upright activity[12]. The tri-axial accelerometer has the additional benefits of detecting the absolute angle of the rear wheel and the direction of movement.

For the purposes of this study, the tri-axial accelerometer was secured to the spokes of a manual wheelchair using a thin sheet of plastic backing and tape or to the inner circumference of a powered wheelchair using double sided tape. The two axes measuring the radial and tangential acceleration components (figure 1), were used to calculate outcome measures in our own analysis algorithm. The third axis, perpendicular to the plane of the wheel, was used to determine when the wheelchair was upright.

FIGURE 1 NEAR HERE

Outcome measures

An algorithm was written using Visual Basic for Applications (Microsoft Corporation, USA), which used the radius of the rear wheel and the radial and tangential components of acceleration to calculate outcome measures of: absolute angle (position of the tri-axial accelerometer on the wheel), wheel revolutions, distance travelled, duration of movement and speed. The raw output from the radial and tangential axes was shifted to be around the zero by subtracting a constant (figure 2a), and the angle of the wheel was calculated from the arctan of the two axes (arctan[radial/tangential]) (figure 2b). Change in angle was calculated as the difference between successive points (figure 2c). Cumulative totals of the absolute value of change in angle were derived to give total wheel revolutions (figure 2d). The wheel was defined to be moving if the change in angle of the wheel was more than zero, and when change in angle was negative the wheel was defined as moving backwards (figure 2e). Movement data were smoothed by converting movement periods of less than one second to stationary periods and vice versa, this was only implemented if the subsequent stationary period was shorter. Duration of movement was derived from the number of movement data points, and speed was derived from the distance travelled and duration of movement. This data processing algorithm was applied to the raw output of the tri-axial activPAL.

FIGURE 2 NEAR HERE

Participants

Fourteen participants with SCI (9 male, 5 female, mean age 37.6 ± 16.7) were recruited from in-patients at the Queen Elizabeth National Spinal Injuries Unit, Glasgow, UK; and were between three months and one year since injury. Participants had a range of injury levels and regularly used a wheelchair for indoor and outdoor use (Table 1). Ethical approval was provided by the South Glasgow and Clyde NHS Ethics Committee and Glasgow Caledonian University School of Health and Social Care Ethics Committee. Participants provided written informed consent prior to commencement of the study.

TABLE 1 NEAR HERE

Experimental Protocol

To assess the concurrent validity the monitoring system was compared with video analysis. Firstly, participants propelled their wheelchair over an indoor circular track at their self-selected speed. Secondly, participants propelled their wheelchair at their self-selected speed on an outdoor wheelchair skills course. All participants used the same tri-axial accelerometer throughout the study. In order to assist video analysis of the absolute angle and wheel revolutions, eight colour-coded plastic markers were attached to the wheel placed at regular intervals. The marker in line with the accelerometer was used to measure absolute angle.

Indoor Protocol

Participants positioned themselves on an indoor circular track and remained stationary in their wheelchair for 30 seconds. Participants then propelled themselves in their wheelchair around a continuous single circuit at a self-selected speed followed by a second stationary period.

Outdoor Protocol

On an outdoor wheelchair skills course incorporating forwards and backwards movement participants performed a variety of activities, ramp manoeuvres with gradients of 5 and 20^{o} (n=2), obstacle manoeuvres with left and right turns (n=2), 3.6m gravel path (n=1) and 100-130mm height kerbs (n=3). Between each section of the course participants were stationary for a short period of time to differentiate between the individual components of each of the eight activities. Figure 2 displays an example of a ramp manoeuvre which was divided into three short standardised sections of forwards and backwards movement separated by two stationary periods. If participants were unable to perform any of the outdoor manoeuvres, for example, as a result of poor upper limb strength or poor back wheel balance, these items were excluded from the protocol.

Video analysis

Throughout the study a hand-held digital video recorder focussed on the participants' right rear wheel. The camera was held perpendicular to the plane of the rear wheel during stationary periods at the start and end point of each activity in order to provide a clear view of the absolute angle with minimal parallax. While the wheelchair was in motion, the rear wheel was kept in the field of view of the camera.

Data analysis

The three outcome measures of absolute angle, wheel revolutions and duration of movement, necessary for measuring speed and distance travelled in the wheelchair, were selected for validation. Absolute angle for the stationary start and finishing point for each bout of movement was analysed using Siliconcoach Pro 7 (Siliconcoach, Otago, New Zealand), a commercially available computer program designed to calculate the angle between markers[14]. The number of revolutions was recorded manually by observation. Incomplete revolutions were quantified by calculating the starting and finishing angle of the activity monitor on the rear wheel. Duration of movement was recorded using the timer on the video by two independent raters, and the average observed time of the two raters was used. Accelerometer data was downloaded to a PC using the activPAL Professional software version 5.8.2.2 (PAL Technologies Ltd, Glasgow, UK). Absolute angle, wheel revolutions and duration of movement were then obtained from the analysis algorithm written by the authors.

Statistical analysis

Concurrent validity of absolute angle, number of wheel revolutions, and duration of movement between the video analysis and monitoring system was assessed using Intraclass-Correlation Coefficients (ICC 2,1) and the Bland Altman method[15] using SPSS version 16.0 (SPSS Inc., Chicago, IL, USA). The agreement between the two independent raters for duration of movement was assessed by the same method.

Results

No data was lost during the study. Only one participant [14] was competent with the kerbs and gravel path, which were included in the analysis. All other participants did not have the skills or upper limb strength to complete these obstacles. Outcome measures were obtained for all individual periods of movement in the protocol, which gave nine data comparisons per person, with an additional four for the participant who completed the kerb and gravel activities. Starting and finishing angles were grouped together for analysis. Table 2 shows a summary of the differences in wheel revolutions, absolute angle and duration for each outdoor activity. The average distance and speed of each activity have been indicated to provide contrast.

TABLE 2 NEAR HERE

Wheel Revolutions

The mean difference of wheel revolutions (video-activity monitor) was 0.002±0.016 (mean±sd) with an absolute maximum difference of 0.038 revolutions (Table 2). The mean absolute percentage error was 0.59% for all tasks, which would correspond to a distance of 0.01m using a regular manual wheelchair with a radius of 0.3m. The activity monitor demonstrated excellent validity (ICC(2,1)= 1.00, 95%CI, 1.00, 1.00) for wheel revolutions. The Bland and Altman method demonstrated an excellent level of agreement with an upper level of agreement (ULOA) of 0.032 and a lower level of agreement (LLOA) of -0.029 (Figure 3a).

FIGURE 3 NEAR HERE

Absolute Angle

The mean difference in absolute angle (video-activity monitor) was 0.006±3.853^o with an absolute maximum difference of 8.789^o (Table 2). The activity monitor demonstrated excellent validity (ICC(2,1)= 0.999, 95%CI 0.999, 0.999) for absolute angle of the activity monitor on the rear wheel and the Bland and Altman method demonstrated an excellent level of agreement with an ULOA and LLOA of 7.545 and -7.558 respectively (Figure 3b).

Duration of movement

The agreement between the two independent raters was excellent with a mean difference of 0.647s (ICC(2,1)= 0.996, 95%CI 0.995, 0.997) The mean difference in duration of movement between the raters and the activity monitor (raters-activity monitor) was -1.868±1.392s with an absolute maximum difference of 7.15s (Table 2). The activity monitor demonstrated excellent validity (ICC(2,1)= 0.981, 95% CI 0.669, 0.994) for duration of movement. The Bland and Altman method demonstrated excellent level of agreement with an ULOA of 0.861 and a LLOA of -4.597 (Figure 3c). Activities ranged from two seconds to one minute in duration and there was a tendency towards wider differences in short movements, with the activity monitor overestimating duration of movement compared with the two independent raters (Figure 3c).

Discussion

The results demonstrate that this novel technique of measuring physical activity performed in the wheelchair using a tri-axial accelerometer and analysis algorithm is valid for measuring wheel revolutions, absolute angle and duration of movement. This monitoring system was valid for activities ranging in distance, duration, speed and direction. From these data it is possible to determine speed and distance travelled, which are central constructs of monitoring activity in wheelchair users. Therefore, this monitoring system can quantify the extent of mobility in a powered wheelchair and can provide an indication of the speed, frequency and duration of the physical activity performed in a manual wheelchair, which can be compared with current physical activity guidelines for the general population[16]. If the monitor is worn for several days these outcome measures have the potential to provide health professionals with an indication of a person's physical activity levels as well as their mobility and integration into the community; this information may be used to evaluate and progress rehabilitation.

The choice of monitoring device greatly depends on the population and the desired aspect of mobility or outcome measure of interest. A monitor positioned on the wheelchair cannot distinguish between self propulsion and being pushed or free-rolling. However it does give a robust measurement of movement performed in the wheelchair, for example, to assess the extent of wheelchair use, mobility and community locomotion in manual or powered wheelchair users, and is unobtrusive as it is not worn on the body and therefore does not need to be removed and reattached. Other methods of monitoring wheelchair activity include placing monitors on the upper limb, which requires an algorithm to separate propulsion from other upper limb tasks; this incurs a risk of false classification[7]. Outcome measures available from upper limb worn activity monitors are energy expenditure, activity counts, and duration of wheelchair propulsion[7, 8, 9]; and small movements or manoeuvring are either not detected or excluded from analysis[7, 10]. The monitoring system used in this study provides outcomes that are easily understood and detects all movements of any magnitude lasting more than one second. These may indicate the accessibility of the environment and the opportunities in daily life for people mobilising using a wheelchair, giving a better understanding of the overall movement patterns regardless of the propulsion technique, direction and speed.

Limitations

The study made use of a small sample of 14 participants with SCI, 11 of whom propelled a manual wheelchair. However, the sample size in this study was comparable to sample sizes in previous similar work[7, 9] and represented a range of abilities and wheelchairs types, travelling at various speeds. The majority of participants were not able to perform the kerbs and gravel path activities. The maximum differences between the monitoring system and video analysis for the participant who did perform these tasks were within the ranges found in the rest of the study, however further work should be conducted to ensure this monitoring system is valid for these tasks. All activities performed during the protocol were of short duration, with the longest activity lasting one minute. People with SCI have been mostly found to continuously propel their wheelchair for 10-30 seconds, and for no more than five minutes[17], so the protocol employed in this study may be regarded as representative of the duration of propulsion activities carried out by the SCI population. The monitoring system showed a tendency to overestimate duration of movement with a maximum difference of 7.15s. This was probably due to some settling motion after the participant came to a halt, resulting in small changes in acceleration which were registered as movement by the device which was continuously recording and sensitive enough to detect this, whilst the raters considered the wheelchair to be stationary. While turning it is possible that both wheels will not move an equal distance. Therefore when the accelerometer is attached to only one wheel the distance moved during turning could be misrepresented. This difference is likely to be small and monitors could be attached to both wheels to overcome this issue.

Conclusion

This study has found the tri-axial activPAL accelerometer placed on the wheelchair rear wheel and newly developed algorithm can accurately measure wheel revolutions, absolute angle, and duration of movement for activities of various distances and duration, during forwards and backwards movement. From these data it is possible to determine the distance, speed and duration of activities performed in the wheelchair.

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Conflict of Interests

One of the authors (MG) is a co-inventor of the activPAL physical activity monitor and a director of PAL Technologies Ltd. However that author was not involved in data collection or the statistical analysis of the results. The remaining authors declare no competing interests.

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Figure 1. Demonstrating the attachment of the tri-axial accelerometer to the rear wheel of a manual wheelchair. Absolute angle of the wheel is measured from 0^o to the centre of the activity monitor.

Figure 2. Graphs showing an excerpt of wheelchair propulsion. a) radial (solid line) and tangential (dashed line) acceleration components of the tri-axial accelerometer, b) absolute angle of the wheel, c) change in angle, d) wheel revolutions of the rear wheel, e) direction and stationary periods of the wheelchair, ADC units= analogue to digital converted units, F= forward movement, S= stationary, B=backward movement.

Figure 3. Bland Altman plots (mean vs difference; video – activity monitor) of all data for: a) wheel revolutions, b) absolute angle, c) duration of movement.

Table 1. Characteristics of Participants.

PARTICIPANT	M/F[1]	AGE	SCI LEVEL	ASIA ISC[2]	MANUAL/ ELECTRIC	WHEELCHAIR TYPE	WHEEL RADIUS (cm)
1	M	18	C4	A	ELECTRIC	INVACARE SPECTRA PLUS	16
2	M	48	C5	A	ELECTRIC	INVACARE SPECTRA PLUS	16
3	F	18	C6	C	MANUAL	KUSCHALL AIRLITE	30
4	M	44	C6	B	MANUAL	QUICKIE ARGON	30
5	F	42	C6	C	MANUAL	KUSCHALL COMPACT	30
6	F	48	C7	D	MANUAL	KUSCHALL COMPACT	30
7	M	60	C8	C	MANUAL	KUSCHALL COMPACT	30
8	M	29	T4	A	MANUAL	QUICKIE ARGON	30
9	M	18	T6	A	MANUAL	KUSCHALL AIRLITE	30
10	F	55	T8	B	MANUAL	KUSCHALL AIRLITE	30
11	F	65	T8	C	ELECTRIC	INVACARE SPECTRA PLUS	16
12	M	39	L1	B	MANUAL	QUICKIE NEON	30
13	M	20	L3	A	MANUAL	KUSCHALL K SERIES	30
14	M	22	L3	D	MANUAL	KUSCHALL AIRLITE	30

[1] M=Male, F=Female

[2] ASIA ISC: American Spinal Injury Association Impairment Scale Classification, A= complete injury, B=sensory function is preserved below the SCI level, C= motor function is preserved below the SCI level, more than half of the key muscles below the level of injury have a muscle grade less than three, D= motor function is preserved below the SCI level, at least half of the key muscles below the level of injury have a muscle grade of three (1).

Table 2. Summarised results

	Difference in Angle (^{o})	Difference in Total Revolutions	Difference in Time (s)	Distance Travelled (m)	Speed (m/s)
Indoor Circular Track (n=14)	0.38 ± 3.08 [-4.47-6.07]	0.009 ± 0.013 [-0.013- 0.025]	-2.20 ± 1.21 [-4.55- 0.050]	21.91	0.001
1st Ramp & Manoeuvre (n=14)	-0.24 ± 3.87 [-7.71-7.23]	0.001 ± 0.014 [-0.032- 0.033]	-1.93 ± 1.56 [-6.25- 0.080]	4.87	0.33
1st Obstacle Manoeuvre (n=14)	1.08 ± 3.53 [-5.42- 6.85]	0.006 ± 0.014 [-0.026- 0.028]	-2.46 ± 1.89 [-7.15- -0.30]	9.58	0.47
2nd Obstacle Manoeuvre (n=14)	-0.96 ± 3.83 [-7.73- 6.09]	0.002 ± 0.018 [-0.032- 0.038]	-1.63 ± 1.74 [-5.90- 2.05]	42.27	1.05
2nd Ramp & Manoeuvre (n=14)	0.055 ± 3.90 [-8.44- 8.43]	0.000 ± 0.016 [-0.030- 0.036]	-1.56 ± 0.97 [-4.10- 0.00]	5.79	0.28
Kerbs & Gravel (n=1)	0.81 ± 4.08 [-3.30- 8.79]	0.010 ± 0.017 [-0.015- 0.025]	-2.08 ± 1.15 [-2.95- -0.40]	3.01	0.27