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Original Article

Accuracy and Precision of Actigraphy and SMARTwheels for Measuring Push Counts Across a Series of Wheelchair Propulsion Trials in Non-disabled Young Adults

Hunter Soleymani,[1] Brenda Jeng,[2] Beshoy Abdelmessih,[1] Rachel Cowan,[3] Robert W. Motl.[2]

Abstract

Background: There has been a growing interest in "Lifestyle Physical Activity" (LPA) among wheelchair users. LPA can be quantified via "pushes" as an outcome metric. This study examined the accuracy and precision of research-grade devices for counting pushes across a series of wheelchair propulsion trials. **Methods:** Eleven non-disabled, young adults completed 19, 1-minute wheelchair propulsion trials at self-selected speeds with a wheelchair equipped with a SMARTwheel (SW) device while being video recorded. Participants also wore 2 ActiGraph accelerometers, one on the wrist and one on the upper arm. Video footage enabled manual counting of the number of pushes (gold standard). Total pushes were averaged across 16 workloads (3 trials of repeated workloads were excluded) for each device and compared to manually counted pushes. **Results:** Compared to manually counted pushes, SW demonstrated the greatest accuracy (mean difference [MD] compared to video of 2.3 pushes [4.5% error]) and precision (standard deviation of the mean difference [SDMD]) compared to video of 4 pushes, (Coefficient of Variation [CV] =.04), followed by the upper arm-worn accelerometer (MD of 4.4 pushes [10.4% error] and SDMD of 10, [CV= .06]) and the wrist-worn accelerometer (MD of 12.6 pushes [27.8% error] and SDMD of 13 [CV=.15]). **Conclusions:** SW demonstrated greater accuracy and precision than ActiGraph accelerometers placed on the upper arm and wrist. The accelerometer placed on the upper arm was more accurate and precise than the accelerometer placed on the wrist. Future investigations should be conducted to identify the source(s) of inaccuracy among wearable push counters.

Key Words: Wheelchair; Actigraphy; Physical Activity; Health Promotion; Disability (Source: MeSH-NLM).

ClinicalTrial.gov identifier. https://clinicaltrials.gov/ct2/show/NCT04987177

Introduction

There has been a growing interest in the study of physical activity for management of health outcomes among wheelchair users and this has largely focused on participation in intentional, structured, and planned exercise training.[1, 2] Nevertheless, there are many barriers for participation in this type of physical activity, and such barriers may underlie the low number of wheelchair users who achieve the recommended physical activity levels.[3-6] To that end, researchers have recently advocated for a paradigm shift towards organic incorporation of health-promoting physical activity into daily life, termed "Lifestyle Physical Activity" (LPA).[1, 5] The paradigm shift advocates for an application of concepts regarding LPA among those who use manual wheelchairs as a primary or only means of mobility (i.e., spinal cord injury, multiple sclerosis, cerebral palsy, and spina bifida). The paradigm shift includs suggestions for a working definition and metrics of LPA for manual wheelchair users followed by a brief discussion of LPA correlates, consequences, interventions, and safe movement considerations.

One of the key steps in meeting the challenges of this paradigm change involves tools for monitoring "pushes" as a metric of LPA. To date, little is known regarding the accuracy and precision of research-grade devices, such as SMARTwheels [SW] and ActiGraph accelerometers, for monitoring pushes as a metric of LPA. Such research is important for documenting changes in LPA pre/post intervention and for better identifying associated outcomes of LPA in wheelchair users. SWs have a long history of providing reliable data and being a critical instrument for wheelchair research studies involving the relationship between the type of wheelchair, set-up, activity, technique, anatomy, physiology, and repetitive strain injury.[7] SW devices are considered the gold standard but are not cost-effective and currently no longer in production (SW cost: $15,000 USD in 2012,

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ActiGraph Accelerometer cost: $430 USD, Apple Watch Series 8 cost: $399 USD, and Fitbit Flex 2 cost: $229 USD). There has been recent interest in the accuracy and/or precision of commercially available wearable devices such as Apple Watch[8-11] and Fitbit.[8] The Apple Watch Series 4 has demonstrated a mean absolute percentage error (MAPE) of 9.2-13.9%[8, 9] compared with manual counting of pushes during wheelchair propulsion, and this was substantially better than the Fitbit Flex 2 (MAPE of 59.7%).[8] To our knowledge, there are currently no data on the accuracy and/or precision of research-grade devices for push counts.

The current paper extends previous research and explores research-grade tools for measuring pushes as an outcome metric of interventions designed for promoting LPA in wheelchair users. If we can provide accurate and precise measurements of pushes, future research can better examine the relationship between physical activity and its correlates in manual wheelchair users, so that clinicians may prescribe, promote, and monitor LPA. Accordingly, we examined the accuracy and precision of ActiGraph accelerometers and SW for measuring push counts during 19 bouts of manual wheelchair propulsion in healthy young adults. We expected that SW would demonstrate greater accuracy and precision than the wearable ActiGraph accelerometers. Additionally, we examined the accuracy and precision of research-grade accelerometers based on location on the arm (i.e., wrist vs. upper arm) and expected that the accelerometer on the upper arm would demonstrate better accuracy and precision for counting pushes than the accelerometer placed on the wrist. This study is a proof-of-concept pilot project conducted between August 2021 and November 2021 during the COVID-19 pandemic. We tested non-disabled individuals to enable a rapid evaluation of the accuracy and precision of research-grade devices. This was necessary as individuals with spinal cord injury, who are commonly enrolled in wheelchair studies, are particularly vulnerable to respiratory infections and other complications.[12-14] We sought to reduce risks of COVID-19 exposure by using non-disabled individuals.

Methods

Participants

This research protocol was approved by the University of Alabama at Birmingham Institutional Review Board (IRB-30007513) and registered with ClinicalTrials.gov (NCT04987177). Eleven non-disabled adults were recruited through local flyers, medical school interest groups, and word of mouth, and all participants provided written consent prior to participation. These data are secondary analyses of a parent study (Clinical trial registration number: NCT04987177). The parent study had 90% power at α=0.05 to detect a repeated measures correlation of 0.238 (two tail) with 12 participants, each completing 16 repeated measures. Our final sample size of n=11 was similar in size to many other wheelchair propulsion studies that enrolled wheelchair users[15-19] or non-disabled individuals.[20-24] Inclusion criteria were (a) age ≥18 years, (b) ability to safely participate in vigorous physical activity (assessed by the Physical Activity Readiness Questionnaire for Everyone [PAR-Q+], and (c) no current usage of a wheelchair. Exclusion criteria were failure to meet all the inclusion criteria. Inclusion and exclusion criteria were selected to maximize the participant safety and protocol completion. No adverse events occurred during testing.

Instrumentation and Configurations

All testing was performed using the same TiLite (TiLite, Permobil, Timra, Sweden) wheelchair (specifications in accordance with the recommendations of Fritsch et al. are in *Supplemental Table 1*).[25] The submaximal peak test was performed with SHOX (Custom Engineered Wheels, Inc., Baldwyn, MS, USA) solid tires mounted to TiLite Shadow 25″ wheels. The within-subject repeated measures protocol was performed with a 25″ Primo (Xiamen Lenco Co, LTD, Xiamen, China) pneumatic tire on the left side and a 25″ SMARTwheel equipped with matching pneumatic tire on the right side. During all testing, the wheelchair was secured to a WheelMill ergometer using two straps attached to the wheelchair backrest stabilizer bar and 1 strap across the foot plate.[26] We manipulated rolling resistance by adjusting the WheelMill parameters of testing decay and force multiplying coefficients,[26] which both are inversely related to rolling resistance (i.e., ↓ decay/force multiplying coefficient = ↑ rolling resistance). Participants were equipped with two ActiGraph GT3X+ accelerometers (ActiGraph, LLC, Pensacola, FL, USA); 1 on the right wrist above the distal radioulnar joint and 1 on the right upper arm at a point halfway between the lateral epicondyle of the elbow and the greater tubercle of the humerus. The accelerometers were calibrated by the manufacturer prior to the start of the study. The accelerometer is a lightweight, small device that contains a solid-state accelerometer that generates an electrical signal proportional to the force acting on it along three axes. Acceleration detection ranged in magnitude from 0.5-2.5g, and the frequency ranged from 0.25-2.50Hz. The signal was digitized by a 12-bit analog converter and integrated over 1s epoch intervals. The data were downloaded via the ActiLife software using a sample frequency of 100Hz and reintegrated into vector magnitude per 1s epoch with the low frequency extension applied and imported to Microsoft Excel for further processing. Vector magnitude was expressed as counts per minute across each bout of manual propulsion. 2D sagittal view video footage was collected from the right side.

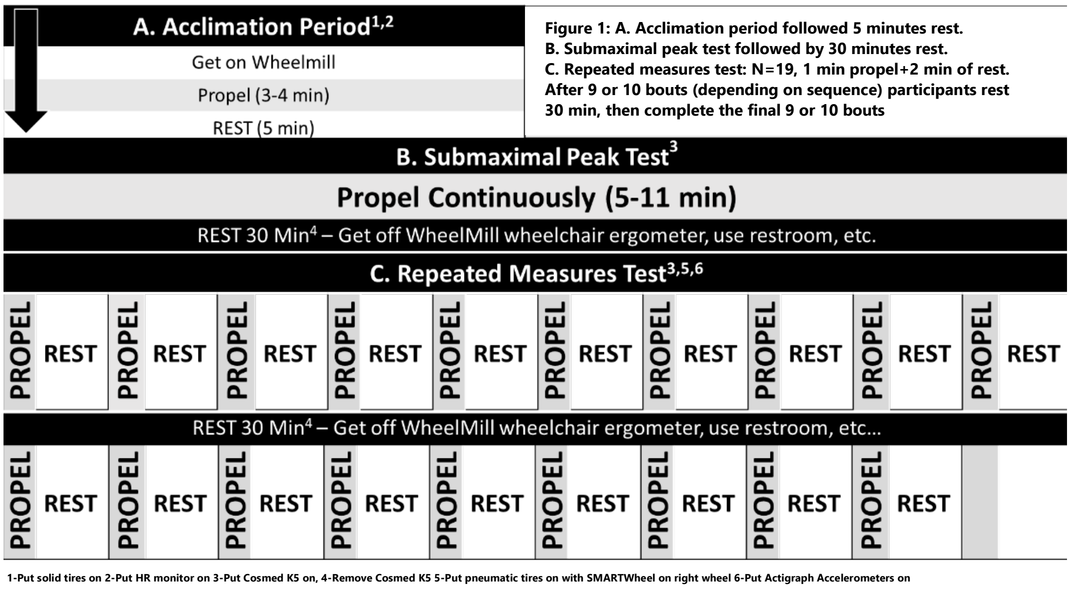
Rating of Perceived Exertion (RPE)

A non-differentiated 0-10 OMNI scale validated for use in manual wheelchair propulsion testing[27] was used to monitor perceived exertion during the acclimation period, submaximal test, and repeated measures protocol. Participants were introduced to the scale during the consent process and refamiliarized with the scale prior to the acclimation period, submaximal test, and repeated measures protocol.

Acclimation Period

A summary of the entire protocol can be found on *Figure 1*. Since participants were non-disabled persons with minimum previous wheelchair propulsion experience, we implemented an acclimation period prior to the graded exercise test and repeated measures protocol. Participants were instructed to "propel at a casual pace that was comfortable for them" for 3-4 minutes. During this time, rolling resistance was manipulated, and RPE27 was collected every 30-45 seconds. Participants were allowed to change pushing speeds as resistances changed to maintain a comfortable pace, and this would naturally change pushing cadence. The starting resistance and resistance changes were based on the teams prior Wheelmill experience. The acclimation period was considered complete once the participant had completed a minimum of three minutes and we had identified at least one resistance rated as "easy" (RPE=2) and at least one rated as "hard" (RPE≥7). The "easy" resistance was used as the beginning resistance for the submaximal test. The speed pushed during the "easy" resistance was used as the target speed participants maintained during the submaximal peak test. We required experience of a "hard" rolling resistance to ensure participants had experienced it prior to the submaximal and repeated measures testing. Participants rested for at least 5 minutes following the acclimation period.

Figure 1. Summary of Testing Protocol.



Data Collection

Submaximal test to estimate maximum workload

The submaximal test estimated the maximum workload for use in the repeated measures protocol. Each participant completed the submaximal graded exercise test at the speed established during the acclimation period. Participants pushed continuously for the entire test, with workload (i.e., rolling resistance) increasing every minute until the participant reached RPE=8. The starting rolling resistance for each participant was established based on acclimation phase where RPE=2 rolling resistance (i.e., the same values for the WheelMill control parameters were input). Rolling resistance was increased each minute by a constant amount (i.e., a 0.04 unit decrease in the WheelMill parameter "force multiplying coefficient"). RPE was documented during the last 20 seconds of each one-minute stage. Participants rested for at least 30 minutes before starting the repeated measures protocol.

Each participant's maximum (i.e., 100%) workload capacity was estimated from the RPE-force multiplying coefficient relationship measured during the submaximal test. Maximum capacity (i.e., 100% workload) was defined as the estimated force multiplying coefficient at RPE=10. For each participant, RPE was regressed on force multiplying coefficient to generate the individualized linear equation of equation 1.

Equation 1:

(RPE × beta) + constant = force multiplying coefficient

RPE=10 was then plugged in to estimate the force multiplying coefficient at maximum capacity (i.e., 100% workload). This estimated force multiplying coefficient was set as the 100% rolling resistance level tested during the repeated measures protocol and was used to generate all other resistance levels tested using equation 2.

Equation 2:

Resistance level = target % × 100% force multiplying coefficient

Within-Subject Repeated Measures Test

Participants next completed a single-blind, within-subject, repeated measures experiment. Each participant completed 19, 1-minute propulsion bouts at a self-selected speed. The 19 bouts

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consisted of 16 unique resistance levels between 25% and 100% in 5% increments of each participants estimated maximum capacity (i.e., 25%, 30%, 35%, etc.). Three resistance levels (25%, 50%, 75%) were completed twice, once in each block. To reduce potential fatigue effects, the 19 trials were divided into two blocks. Block 1 included 9 trials and block 2 included 10 trials. The trials were partitioned in a manner that total workload, defined as the sum of the resistance levels (% max), was equal between blocks. Within each block, trial order was designed to have an unpredictable pattern of increases/decreases in resistance and featured the highest rolling resistance trials towards the middle of the set. Participants completed the blocks in a counterbalanced order within gender (*Table 1*). Participants rested for 2 minutes after each one-minute trial and rested for 30 minutes between blocks. An automatic timer with a bell was used to instruct the participants when to begin and end each trial. Heart rate was recorded at the 40-second mark of each trial, and RPE was recorded immediately following the end of each trial.

Video Counting Process

Videos of each one-minute trial were deidentified, randomized, and divided into four batches for counting. Each one-minute clip was viewed by one person. A stroke count was recorded using a tap counter application using the following criterion: A stroke was counted at the end of each cycle after the subject touched the wheel, pushed forward, and then let go. Each batch was counted twice before moving onto the next batch (i.e., batch 1 counted twice, then batch 2 counted twice, etc.). Once the count was completed, the results were recorded into a spreadsheet, and any discrepancy was recorded and discussed.

Statistical Analysis

Data analyses were conducted for n=16 trials (the second trial for the 25/50/75% conditions were not analyzed) in SPSS version 28 (IBM, SPSS Inc., Chicago, IL). We evaluated accuracy and precision with absolute and relative metrics. Absolute accuracy was calculated as the mean difference between manually counted pushes and device-measured pushes. Relative accuracy was assessed as percentage error (i.e., [mean difference between manually counted pushes and device-measured pushes ÷ by manual pushes] × 100) and the frequency of large errors per device was based on ≥5%, ≥10%, and ≥25% error. Absolute precision was assessed as the standard deviation of the mean difference, and relative precision was assessed as the coefficient of variation (CV). We provided Bland-Altman plots to illustrate metrics of absolute accuracy and relative precision. We further conducted Spearman rho's bivariate correlation analyses among manually recorded push count difference, workload, rolling resistance, power output, and speed to evaluate sources of inaccuracy in counting pushes among ActiGraph accelerometers.

Results

Participants

Eleven (7 males, 4 females) non-disabled individuals with minimal previous experience propelling a manual wheelchair completed the study. Mean age (SD) was 24 years (+/-2.3 y), ranging from 22 to 29. Based on body mass index (BMI), 8 participants were normal weight (18.5-24.9 kg/m2), 1 was overweight (25-29.9 kg/m^2), and 2 were obese (≥30 kg/m^2) (*Table 1*).

Table 1. Participant Characteristics of the Sample of Non-disabled Young Adults (n=11).

Participant Number	Gender	Age (years)	Race/Ethnicity	Height (cm)	Weight (kg)	BMI (kg/m^2)	Sequence
	M	29	White	183	77.1	23.1	B
2	F	23	White	163	87.1	33.0	B
3	M	23	White	178	77.7	24.6	A
4	M	22	White	188	74.8	21.2	B
5	F	28	White	168	52.7	18.8	A
6	M	23	White	173	77.8	26.1	A
7	F	24	White	168	54.1	19.3	B
8	M	22	White/Asian	180	73.0	22.5	B
9	M	22	Asian/Hispanic	175	70.1	22.8	A
10	M	24	White	191	111.5	30.7	B
11	F	22	White	170	59.0	20.4	A
Average / Total	M=7 F=4	24±2.33	White only=9 All Other=2	176 ±8.43	74.1 ±15.69	23.8 ±4.32	A=5 B=6

Legend: Data are presented as number or mean +/- SD. M Male; F Female. Sequence A was block X, 30 min rest, block Y. Sequence B was block Y, 30 min rest, block X. Block X trial order (n=9, sum=575%): 55%, 50%, 70%, 75%, 100%, 90%, 25%, 30%, 80%. Block Y trial order (n=10, % sum=575%): 25%, 50%, 35%, 95%, 85%, 65%, 45%, 40%, 75%, 60%.

Accuracy

Metrics for absolute and relative accuracy are presented in *Table 2* and illustrated in *Figures 2-5*. Push counts captured by the wrist ActiGraph deviated from the manually counted condition by a mean of 12.6 (27.8% error) pushes. The frequency of small (≥5% error), medium (≥10% error), and large (≥25% error) errors were 115 (66%), 98 (56%), and 79 (45%), respectively. Push counts captured by the upper arm ActiGraph deviated from the manually counted condition by a mean of 4.4 (10.4% error) pushes. The frequency of small (≥5% error), medium (≥10% error), and large (≥25% error) errors were 44 (25%), 34 (19%), and 25 (14%), respectively. Push counts captured by the SW deviated from the manually counted condition by a mean of 2.3 (4.5% error) pushes. The frequency of small (≥5% error), medium (≥10% error), and large (≥25% error) errors were 25 (14%), 23 (13%), and 13 (7%), respectively.

Precision

Metrics for absolute and relative precision are presented in *Table 3* and illustrated in *Figures 2-5*. Regarding the wrist ActiGraph, the SD of the mean difference compared with video was 13 (CV=.15). Regarding the upper arm ActiGraph, the SD of the mean difference compared with video was 10 (CV=.06), whereas the SD of the mean difference for the SW compared with video was 4 (CV=.04).

Spearman's Rho correlations

Spearman's rho correlations between upper arm ActiGraph-Video push count difference and workload, rolling resistance, power output, and speed are provided in *Table 4*. Upper arm ActiGraph-Video push count difference was significantly associated with rolling resistance (ρ=-0.174, p=0.022) and power output (ρ=-0.268, p<0.001). However, upper arm ActiGraph-Video push count difference were not associated with workload (ρ=-0.070, p=0.354) and speed (ρ=-0.137, p=0.072).

The study examined the accuracy and precision of the ActiGraph accelerometers and SWs for measuring push counts during manual wheelchair propulsion. The SW provided more accurate and precise estimates of push counts compared with accelerometers placed on the upper arm and wrist. The results further indicated more accuracy and precision of push count measurements with the accelerometer placed on the upper arm compared with the wrist. This preliminary study supports the accuracy and precision of SWs and perhaps upper arm-worn ActiGraph as research-grade devices for quantifying pushes as a metrics of LPA in persons who use manual wheelchairs.

Table 2. Accuracy of ActiGraph GT3X+ Devices Worn on the Wrist and Upper Arm and SMARTWheel for Capturing Pushes During Manual Wheelchair Propulsion Across 16 Trials of Increasing Workloads in a Sample of 11 Non-disabled Young Persons.

	Absolute Accuracy		Relative Accuracy			
	Mean (SD) of Total Pushes Averaged Across 16 Workloads	Mean Difference in Total Pushes Averaged Across 16 Workloads Compared with Video	Mean (SD) Percentage Error	n≥5% error (%)	n≥10% error (%)	n≥25% error (%)
Manually Counted	50(8)					
Wrist ActiGraph	63(12)	12.6	27.8(30.0)	115(66%)	98(56%)	79(45%)
Upper Arm ActiGraph	54(11)	4.4	10.4(24.8)	44(25%)	34(19%)	25(14%)
SMARTwheel	48(8)	2.3	4.5(8.8)	25(14)	23(13%)	13(7%)

Legend: SD standard deviation.

Figure 2. Bland-Altman Plot for Video 2. Negative Y-axis Values Indicate the 2nd Manual Push Counts Were Greater than the 1st Manual Push Count and Vice-versa.

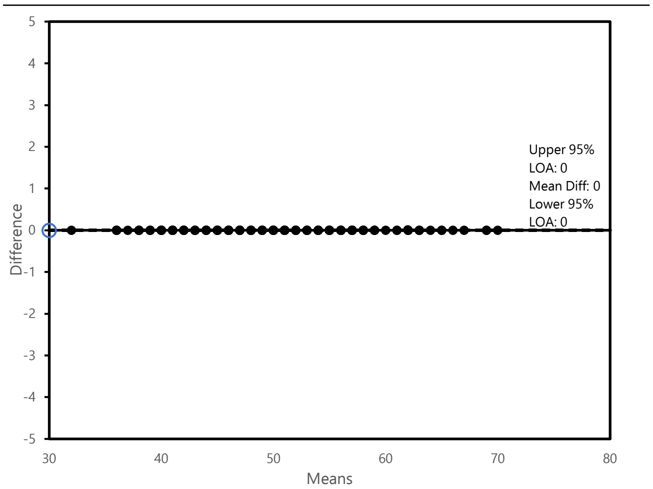


Figure 3. Bland-Altman Plot for the SW. Positive Y-axis Values Indicate SW Push Counts that Were Less Than Manual Push Counts and Vice-versa.

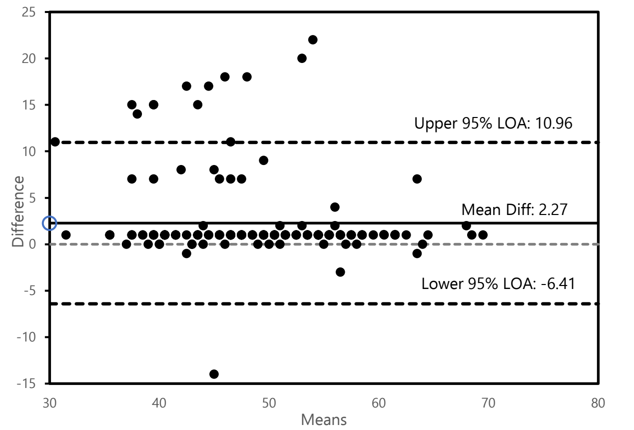


Figure 4. Bland-Altman Plot for the Upper Arm ActiGraph Accelerometer. Positive Y-axis Values Indicate ActiGraph Upper Arm Push Counts that Were Less Than Manual Push Counts and Vice-versa.

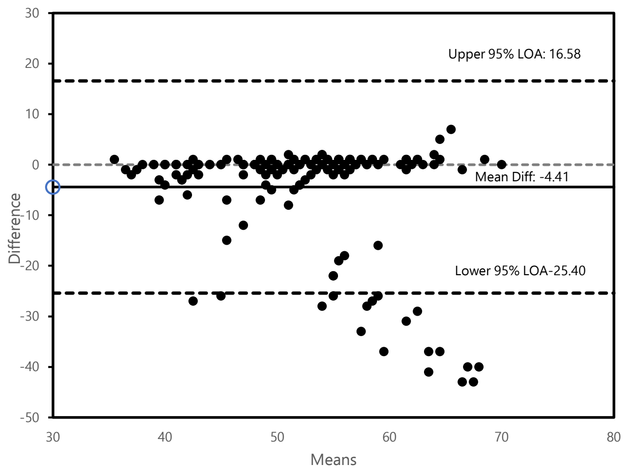
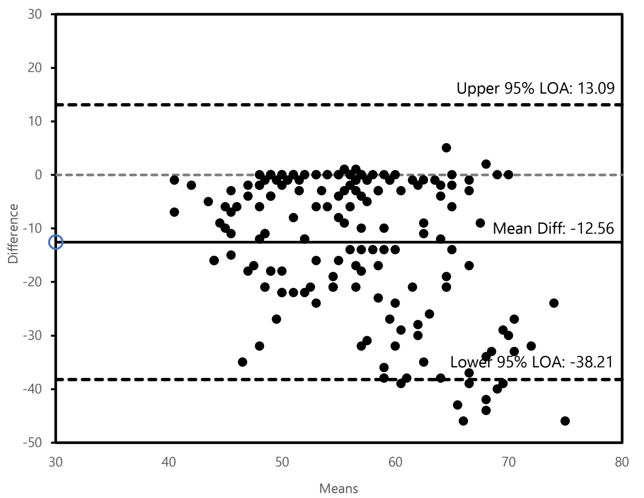


Figure 5. Bland-Altman Plot for the Wrist ActiGraph Accelerometer. Positive Y-axis Values Indicate ActiGraph Wrist Push Counts that Were Less Than Manual Push Counts and Vice-versa.



Discussion

Overall, compared to manual counting, SW slightly undercounted total pushes (SD) averaged across 16 workloads (manual: 50[8] pushes vs SW: 48[8] pushes). We suspect the SW undercounting could stem from discrepancies of defining a "push" or due to a push occurring on the wheel and not the push rim where the sensor on the SW is located. This could be the focus of future research examining the accuracy and precision of SW for measuring pushes in manual wheelchair users.

Conversely, compared to manual counting, both ActiGraph accelerometers overcounted total pushes (SD) averaged across all 16 workloads (upper arm: 54[11] pushes, wrist: 63[12] pushes, manual: 50[8] pushes). Due to limited research in using wearable devices for wheelchair push counts, comparisons of our study population with existing research are limited. Our finding of wearable push counters having the tendency to overcount is somewhat consistent with previous studies evaluating Apple Watch accuracy for counting pushes during wheelchair propulsion.[8-10] However, we identified one study that reported undercounting from the series 1 Apple Watch compared with manual counting during wheelchair propulsion through a 21-part obstacle course.[11] This may be due to differences in the definition of a "push" or in the methodology. For example, one group of researchers[11] defined a push as "any force that was applied to the rim of the wheel by the hand that resulted in movement of the manual wheelchair," including backwards pushes, and the testing protocol included multidirectional/backwards propulsion, whereas our protocol included only forward propulsion. Overall, this suggests that wearable device-measures of push counters tend to overcount during forward wheelchair propulsion. Further investigation is required to evaluate the accuracy and precision of wearable device-measures of push counts during backward wheelchair propulsion.

Table 3. Precision of ActiGraph GT3X+ Devices Worn on the Wrist and Upper Arm and SMARTWheel for Capturing Pushes During Manual Wheelchair Propulsion Across 16 Trials of Increasing Workloads in a Sample of 11 Young Persons.

	Absolute Precision	Relative Precision
	SD of the Mean Difference in Total Pushes Averaged Across 16 Workloads Compared with Video	Coefficient of Variation
Wrist ActiGraph	13	.15
Upper Arm ActiGraph	10	.06
SMARTwheel	4	.04

Legend: *SD* standard deviation.

The tendency for wearable push counters to overestimate can possibly be explained by increased "noisiness" of hand/arm motion during a push, resulting in falsely counted pushes. Based on *Figure 3*, for a large portion of the time, the upper arm ActiGraph accelerometer was accurate, but there was a subset of trials in which the accelerometer push counts varied significantly from the manually recorded pushes counts (the gold standard). We evaluated hand-traced patterns during the wheelchair propulsion to determine if certain motions/hand patterns (i.e., vertical hand accelerations inherent in some certain push pattern trajectories) contributed to the inaccuracy of push counts recorded by accelerometers. However, we were not able to confirm this theory. Additionally, we evaluated bivariate correlations between upper arm ActiGraph-Video push count difference and workload, rolling resistance, power output, and speed. Our results suggest that rolling resistance and power output may have influenced the differences between the upper arm worn ActiGraph accelerometer and manually counted pushes. This warrants further investigations of whether or not vertical acceleration or other potential factors (i.e., wheelchair configuration, propulsion mechanics, individual factors) may contribute to these discrepancies in recorded push counts.

Table 4. Spearman's Rho Correlations Between Upper Arm ActiGraph-Video Push Count Difference and Workload, Rolling Resistance, Power Output, and Speed.

	Workload (%)	Rolling Resistance (N)	Power output (W)	Speed (m/s)
(n=11 participants)	-0.070 P=0.354 n=175	-0.174 P=0.022 n=175	-0.268 P<0.001 n=175	-0.137 P=0.072 n=175

Our results suggest that an ActiGraph accelerometer on the upper arm during wheelchair propulsion was more accurate (% error=10.4 vs 27.8) and precise (CV=.06 vs .15) than a unit worn on the wrist for measuring push counts. This further supports our suggestion that increased "noisiness" in arm/wrist motion is a contributing factor of overcounting. During wheelchair propulsion, the activity of the hand/wrist is higher and more variable than the mid humerus portion of the arm. Further work needs to be done to confirm if this pattern is present among more experienced wheelchair users.

Our results suggest that SW (4.5% error) was more accurate than the wrist-worn ActiGraph accelerometer (27.8% error) and an upper arm-worn ActiGraph accelerometer (10.4% error) in our sample of non-disabled young adults. Previous studies have reported series 4 Apple Watch to have an accuracy (9.2-13.9% error),[8 9] which is comparable to the accuracy of our upper arm-worn accelerometer. However, the Apple Watch from the aforementioned study may be more accurate in measuring push counts than the wrist-worn accelerometer in our study. This is contradictory to what one would expect, as ActiGraph is a research-grade device while the Apple Watch is not. Future investigations are needed to identify the source(s) of inaccuracy among wearable push counters and to compare research grade devices to commercially available devices.

Some limitations should be considered when evaluating the results of this study. We included a relatively small sample size of persons who were inexperienced with manual wheelchair propulsion. Future research may include a larger sample size of persons who use manual wheelchairs regularly (i.e., more than 50% of their daily life). Another limitation was that ActiGraph accelerometers were placed only on the right side, as there may differences in push counts between the dominant and non-dominant sides. Furthermore, we used a WheelMill ergometer rather than over-ground manual wheelchair propulsion for this study protocol. Wheelchair propulsion over-ground may have different biomechanical characteristics compared with wheelchair propulsion on an ergometer and may translate to daily life more readily. Another limitation is the use of research-grade devices to capture push counts. A potential avenue of research would be to compare accuracy and precision of commercially available activity monitors for measuring pushes in manual wheelchair users.

Conclusion

This study examined the accuracy and precision of ActiGraph accelerometers and SW for measuring pushes in non-disabled young adults. SWs demonstrated greater accuracy and precision than ActiGraph accelerometers placed on the upper arm and wrist, yet the accelerometer placed on the upper arm was more accurate and precise than the accelerometer placed on the wrist. An area for future investigation includes direct comparison of the accuracy and precision of available wearable devices, including ActiGraph accelerometers, Apple Watch, and Fitbit devices for manual wheelchair push counting. Once the most accurate and precise device is identified and deemed to yield acceptable data, future studies can then focus on furthering our understanding of physical activity and its correlates and consequences in manual wheelchair users. One potential example, among many, includes evaluating the relationship between daily push counts and health outcomes such as cardiovascular disease in wheelchair users.

Summary – Accelerating Translation

Title: Accuracy and Precision of Actigraphy and SMARTwheels for Measuring Push Counts Across a Series of Wheelchair Propulsion Trials in Non-disabled Young Adults

Main Problem to Solve: There has been a growing interest in the study of physical activity for management of health outcomes among wheelchair users. One key step in monitoring physical activity levels involves having tools for monitoring "pushes." To date, little is known about how well research-grade devices work for monitoring pushes. If we can provide accurate and precise measurements of pushes, future research can better examine physical activity among manual wheelchair users, so that clinicians may prescribe, promote, and monitor physical activity.

Aim of Study: Examine the accuracy and precision of SW and ActiGraph accelerometers for measuring push counts during 19, 1-minute bouts of manual wheelchair propulsion in healthy non-disabled adults.

Methods: Eleven (7 males, 4 females) non-disabled, young adults completed the protocol. All testing took place on a wheelchair machine that allowed us to control the resistance they pushed against. The same wheelchair was used for each participant, equipped with a device that counts pushes. Participants further wore 2 devices, one on the wrist and one on the upper arm that counted pushes. Video footage was recorded, which enabled manual counting of the number of pushes (gold standard). Participants underwent an acclimation period to get used to pushing a wheelchair. Then participants underwent an exercise test in which they pushed continuously for 5-10 minutes as the resistance they pushed against increased. Lastly, participants underwent 19, 1-minute pushing bouts against various resistances ranging from 25-100% of the estimated maximum resistance they could push against. We used the data obtained from the device on the wheel, the two devices on the participants arms, and the data from the video recordings to compare how accurate and precise each tool was for counting pushes. The manual counts from the video data were used as the gold standard and is what the other devices were compared to. We also evaluated various push mechanics to see if any certain factor may have caused the devices to count incorrectly.

Results: The device on the wheelchair most the most accurate and precise tool, followed by the device on the participants upper arm, followed by the device on the participants wrist. The device on the wheelchair tended to slightly undercount, while both devices on the participants arms tended to overcount. We were not able to identify a particular pattern of pushing that could be responsible for miscounting by the devices, but our results suggest that two push mechanical factors may be associated with miscounting by devices.

Conclusion: Among the three devices we evaluated, the device on the wheelchair is a better tool to use for counting pushes in manual wheelchair propulsion, followed by the device worn on the upper arm, and the device worn on the wrist. Further research needs to investigate potential factors that cause the devices to miscount. Once this is better understood, researchers can better examine physical activity among manual wheelchair users, so that clinicians may prescribe, promote, and monitor physical activity.

References

1. Cowan RE, Silveira SL, Helle T, Laessoe U, Goeg KR, Bangshaab J, et al. Lifestyle physical activity in manual wheelchair users - an overlooked public health opportunity. Spinal Cord. 2022;60(2):190-2.
2. Piercy KL, Troiano RP, Ballard RM, Carlson SA, Fulton JE, Galuska DA, et al. The Physical Activity Guidelines for Americans. JAMA. 2018;320(19):2020-8.
3. Klaren RE, Motl RW, Dlugonski D, Sandroff BM, Pilutti LA. Objectively quantified physical activity in persons with multiple sclerosis. Arch Phys Med Rehabil. 2013;94(12):2342-8.
4. Motl RW, McAuley E, Snook EM. Physical activity and multiple sclerosis: a meta-analysis. Mult Scler. 2005;11(4):459-63.
5. Motl RW. Lifestyle physical activity in persons with multiple sclerosis: the new kid on the MS block. Mult Scler. 2014;20(8):1025-9.
6. Buchholz AC, McGillivray CF, Pencharz PB. Physical activity levels are low in free-living adults with chronic paraplegia. Obes Res. 2003;11(4):563-70.
7. Cooper RA. SMARTWheel: From concept to clinical practice. Prosthet Orthot Int. 2009;33(3):198-209.
8. Benning NH, Knaup, Petra, Rupp, Rudiger. Comparison of accuracy of activity measurements with wearable activity trackers in wheelchair users: a preliminary evaluation. GMS Medizinische Informatik, Biometrie und Epidemiologie. 2020;16(2).
9. Benning NH, Knaup P, Rupp R. Measurement Performance of Activity Measurements with Newer Generation of Apple Watch in Wheelchair Users with Spinal Cord Injury. Methods Inf Med. 2021;60(S 02):e103-e10.

10. Glasheen E, Domingo A, Kressler J. Accuracy of Apple Watch fitness tracker for wheelchair use varies according to movement frequency and task. Ann Phys Rehabil Med. 2021;64(1):101382.
11. Karinharju KS, Boughey AM, Tweedy SM, Clanchy KM, Trost SG, Gomersall SR. Validity of the Apple Watch((R)) for monitoring push counts in people using manual wheelchairs. J Spinal Cord Med. 2021;44(2):212-20.
12. DeVivo MJ, Krause JS, Lammertse DP. Recent trends in mortality and causes of death among persons with spinal cord injury. Arch Phys Med Rehabil. 1999;80(11):1411-9.
13. Lemons VR, Wagner FC, Jr. Respiratory complications after cervical spinal cord injury. Spine (Phila Pa 1976). 1994;19(20):2315-20.
14. Tollefsen E, Fondenes O. Respiratory complications associated with spinal cord injury. Tidsskr Nor Laegeforen. 2012;132(9):1111-4.
15. Veeger HE, Meershoek LS, van der Woude LH, Langenhoff JM. Wrist motion in handrim wheelchair propulsion. J Rehabil Res Dev. 1998;35(3):305-13.
16. Koontz AM, Cooper RA, Boninger ML, Yang Y, Impink BG, van der Woude LH. A kinetic analysis of manual wheelchair propulsion during start-up on select indoor and outdoor surfaces. J Rehabil Res Dev. 2005;42(4):447-58.
17. Rammer JR, Krzak JJ, Slavens BA, Winters JM, Riedel SA, Harris GF. Considering Propulsion Pattern in Therapeutic Outcomes for Children Who Use Manual Wheelchairs. Pediatr Phys Ther. 2019;31(4):360-8.
18. Rankin JW, Kwarciak AM, Richter WM, Neptune RR. The influence of wheelchair propulsion technique on upper extremity muscle demand: a simulation study. Clin Biomech (Bristol, Avon). 2012;27(9):879-86.
19. Sanderson DJ, Sommer HJ, 3rd. Kinematic features of wheelchair propulsion. J Biomech. 1985;18(6):423-9.
20. de Groot S, Vegter RJ, van der Woude LH. Effect of wheelchair mass, tire type and tire pressure on physical strain and wheelchair propulsion technique. Med Eng Phys. 2013;35(10):1476-82.
21. van Drongelen S, Arnet U, Veeger DH, van der Woude LH. Effect of workload setting on propulsion technique in handrim wheelchair propulsion. Med Eng Phys. 2013;35(3):283-8.
22. Veeger HE, van der Woude LH, Rozendal RH. Load on the upper extremity in manual wheelchair propulsion. J Electromyogr Kinesiol. 1991;1(4):270-80.
23. Veeger HE, van der Woude LH, Rozendal RH. Effect of handrim velocity on mechanical efficiency in wheelchair propulsion. Med Sci Sports Exerc. 1992;24(1):100-7.
24. Bertolaccini GDS, Carvalho Filho IFP, Christofoletti G, Paschoarelli LC, Medola FO. The influence of axle position and the use of accessories on the activity of upper limb muscles during manual wheelchair propulsion. Int J Occup Saf Ergon. 2018;24(2):311-5.
25. Fritsch C. How Was Studied the Effect of Manual Wheelchair Configuration on Propulsion Biomechanics: A Systematic Review on Methodologies. Frontiers in Rehabilitation Services. 2022;3:863113.
26. Klaesner J, Morgan KA, Gray DB. The development of an instrumented wheelchair propulsion testing and training device. Assist Technol. 2014;26(1):24-32.
27. Gauthier C, Grangeon M, Ananos L, Brosseau R, Gagnon DH. Quantifying cardiorespiratory responses resulting from speed and slope increments during motorized treadmill propulsion among manual wheelchair users. Ann Phys Rehabil Med. 2017;60(5):281-8.

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Supplementary Material

Supplemental Table 1. Wheelchair specifications in accordance with the recommendations of Fritsch et al.[15]

Wheelchair Specification	Measurement
Rear wheel diameter solid tire	25″
Rear wheel diameter pneumatic tire	25″
Rear wheel camber with solid tires	11°
Rear wheel camber with pneumatic tires	8°
Handrim diameter for solid tire	21.5″
Handrim diameter for pneumatic tire	22″
Caster diameter	4″
Seat width x length	18″x18″
Seat height with solid tires	19″
Seat height with pneumatic tires	18.5″
Seat angle	1.5°
Backrest height	9″
Backrest angle	80°
Footrest size	6″ x 9″
From bottom of chair to footrest length	13″
Footrest angle	96°
Back of the seat fore-aft position with respect to the rear wheel axle	5″
Back seat height with respect to the ground with solid tire	28″
Back seat height with respect to the ground with pneumatic tire	28.5″
Fork axis angle	45°