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Predicting Physical Activity Energy Expenditure in Manual Wheelchair Users

TOM EDWARD NIGHTINGALE, JEAN-PHILIPPE WALHIM, DYLAN THOMPSON, and JAMES L. J. BILZON

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ABSTRACT

NIGHTINGALE, T. E., J.-P. WALHIM, D. THOMPSON, and J. L. J. BILZON. Predicting Physical Activity Energy Expenditure in Manual Wheelchair Users. *Med. Sci. Sports Exerc.*, Vol. 46, No. 9, pp. 1849–1858, 2014. **Purpose:** This study aimed to assess the influence of anatomical placement of an accelerometer on physical activity energy expenditure prediction in manual wheelchair users. **Methods:** Ten accelerometer units (ActiGraph GT3X+) were attached to a multiaxis shaker table and subjected to a sinusoidal oscillation procedure to assess mechanical validity and reliability. Fifteen manual wheelchair users (mean $\pm$ SD: age, 36 ± 11 yr; body mass, 70 ± 12 kg) then completed five activities, including desk work and wheelchair propulsion (2, 4, 6, and 8 km·h⁻¹). Expired gases were collected throughout. GT3X+ accelerometers were worn on the right wrist, upper arm, and waist. The relations between physical activity counts and metabolic rate were subsequently assessed, and bias $\pm$ 95% limits of agreement was calculated. **Results:** During mechanical testing, coefficients of variation ranged from 0.2% to 4.7% (intraunit) and 0.9% to 5.2% (interunit) in all axes. During human exercise testing, physical activity counts at each anatomical location was significantly ($P < 0.01$) correlated with metabolic rate (wrist, $r = 0.93$; upper arm, $r = 0.87$; waist, $r = 0.73$). The SEE for each correlation were 3.34, 4.38, and 6.07 kJ·min⁻¹ for the wrist, upper arm, and waist, respectively. The absolute bias $\pm$ 95% limits of agreement values were 0.0 ± 6.5 kJ·min⁻¹, 0.0 ± 8.5 kJ·min⁻¹, and 0.0 ± 11.8 kJ·min⁻¹ for the wrist, upper arm, and waist, respectively. **Conclusions:** The ActiGraph GT3X+ is a reliable tool for determining mechanical movements within the physiological range of human movement. Of the three anatomical locations considered, a wrist-mounted accelerometer explains more of the variance and results in the lowest random error when predicting physical activity energy expenditure in manual wheelchair users. **Key Words:** ACCELEROMETER, EXERCISE, PHYSICAL ACTIVITY, DISABILITY, SPINAL CORD INJURY

The effects of regular physical activity (PA) on the health and well-being of persons with a spinal cord injury (SCI) remain poorly characterized. This is despite cardiovascular disease (CVD) now being the leading cause of mortality in individuals with an SCI and occurring earlier in the lifespan in comparison with able-bodied controls (13). Individuals with an SCI show an abundance of elevated CVD risk factors in comparison with those in matched able-bodied counterparts (2,23). The positive contribution of regular PA on these CVD risk factors and on the maintenance of weight balance is well documented and broadly accepted in ambulatory individuals (21). Results of self-reported PA monitoring in individuals with an SCI (14) suggest that reduced PA may play a role in the progression of these risk factors. However, little is known regarding specific components or patterns of PA that are required to

derive protection from chronic diseases and improve metabolic health in manual wheelchair users (MWU). Therefore, objective measures of PA are required to inform future research efforts and this broader health agenda.

Free-living PA is inherently difficult to measure with precision. This becomes even more problematic within a heterogeneous group such as MWU where, despite movement being restricted to the upper body, differential levels/completeness of SCI lesions result in highly variable movement patterns. Improved assessment of habitual PA would permit appropriate cross-sectional comparisons, allow researchers to comment on the efficacy of behavior change interventions, and potentially inform PA guidelines (4). It has been suggested that self-reported measures are unable to adequately quantify the lower end of the PA continuum (30), and the content of questionnaires adopted previously fails to capture activities specific to the lifestyle of MWU. Self-reported measures, although practical for use in large-scale epidemiological studies, often lend themselves to recall bias, floor effects (lowest score is too high for inactive respondents), and participant over-reporting (28). Considering these limitations and the impracticality of direct observations and indirect calorimetry during free-living assessment, other unobtrusive objective measurement tools that can be used to characterize the association with PA and metabolic health, particularly among cohorts where these conditions are more prevalent, are needed.

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Accelerometers are commonly used to quantify free-living PA (25). Over the past decade, ActiGraph™ has released several models, including the most recent GT3X+, with higher-resolution signals, greater data storage capacity, and increased battery life (19). This model remains to be validated in specific cohorts with differing movement patterns, such as MWU. The first logical step in the development of an objective accelerometer-based tool to assess PA energy expenditure (PAEE) in MWU is to assess its basic mechanical reliability. Various mechanical apparatus have been used to assess the reliability of previous generations of the ActiGraph, such as turntables (24) and rotating wheel setups (5) for the initial computer science and applications (CSA) model and hydraulic shaker tables for newer generations (11,29,31). These are advantageous because investigators can maintain precise control over experimental conditions and simultaneously expose multiple monitors to a wide range of accelerations. Therefore, any variability is exclusively intrinsic to the accelerometer (11) and researchers can shift their attention to identifying and minimizing biological variation such as anatomical positioning.

Multisensor PA monitoring devices, which combine accelerometer(s) with other physiological measurements such as HR (Actiheart; Cambridge Neurotechnology Ltd., Papworth, United Kingdom) or temperature (SenseWear Armband; Bodymedia, Inc., Pittsburgh, PA), can offer improved sensitivity and accuracy of PAEE measurement in ambulatory cohorts (4,8). The validity of combined HR and movement sensor PAEE prediction in MWU remains to be established. Its use may also be heavily reliant on individual calibration, as a result of the high variability in cardiovascular responses to exercise in individuals with differing levels/completeness of SCI (12). Initial research using the SenseWear device revealed considerable energy expenditure estimation errors up to 125.8% (16). It is unclear whether incorporating physiological responses can offer a noticeable improvement in the prediction of PAEE in MWU above and beyond determining the most sensitive anatomical location of a simple triaxial accelerometer in this cohort. Waist-mounted monitors have been shown to underestimate energy expenditure by 24% in MWU with an SCI (15). It is not surprising that manufacturers' energy expenditure prediction algorithms developed on the basis of activity counts generated at the waist during ambulation are unsuitable to derive PAEE of MWU. When Hiremath and Ding (17) examined the correlations between raw activity counts from an RT3 triaxial accelerometer and criterion energy expenditure measured by a portable metabolic cart, the counts on the upper arm demonstrated a better correlation ($R^2 = 0.70$ vs 0.44) with the criterion energy expenditure compared with that in the waist. This observation identifies the arm as a potential location to yield better prediction accuracy and reduced error and also highlights the need for the development of specific algorithms to predict PAEE in MWU.

To our knowledge, there are no published studies on the influence of anatomical placement on the validity of a

GT3X+ accelerometer to determine PAEE in MWU. The aims of this study were twofold. First, this study aimed to assess the validity and reliability of the GT3X+ accelerometer during mechanical testing along each orthogonal axis within the physiological range of human movement. Second, this study aimed to evaluate the effect of anatomical positioning of the GT3X+ accelerometer on the relation between PA counts (PAC) and criterion PAEE during a range of representative activities in MWU.

METHODS

Accelerometer

The GT3X+ activity monitor (ActiGraph, Pensacola, FL) records time-varying accelerations within the dynamic range of $\pm 6g$ and contains a solid-state triaxial accelerometer sensitive to movement along three axes: anteroposterior (X), mediolateral (Y), and vertical (Z). The GT3X+ activity monitor is compact (dimensions, $4.6 \times 3.3 \times 1.9$ cm), lightweight (19 g), and can easily be worn at multiple locations on the body. Each unit is powered by a rechargeable lithium ion battery and has a memory of 512 MB. Approximately 40 d of PA data can be recorded when sampling at a frequency of 30 Hz, although the battery would need recharging after 30 d. To quantify the amount and frequency of human movement, accelerometer outputs are digitized via a 12-bit analog-to-digital converter and passed through ActiGraph's proprietary digital filtering algorithms. To eliminate any acceleration noise outside the normal human activity frequency, digitized signals pass through low-bandwidth (0.25 Hz) and high-bandwidth (2.5 Hz) filters (19). The GT3X+ records time-varying accelerations at a user-defined sampling frequency ranging from 30 to 100 Hz. These are then converted to arbitrary units called "physical activity counts". These are calculated through summing the change in raw acceleration values measured during a specific interval of time or "epoch". Unlike previous models such as the GT1M, the desired epoch length can be selected by the end user (1–240 s) after, rather than before, data collection.

Multiaxis Shaker Table

All reliability testing was conducted using a multiaxis shaker table (MAST-9720; Instron Structural Testing Systems Ltd., High Wycombe, United Kingdom). The MAST-9720 is powered via three vertical, one horizontal, and two lateral hydraulic actuators and is calibrated regularly to an accuracy of 0.1g (Fig. 1).

Experiment 1—Mechanical Testing

The MAST testing conditions were restricted by the maximum displacement amplitude of the horizontal actuator (approximately 62.5 mm), which limited maximum acceleration to 1.5g. With the limitations of the MAST rig, a similar testing schedule to that used by Horner et al. (16)

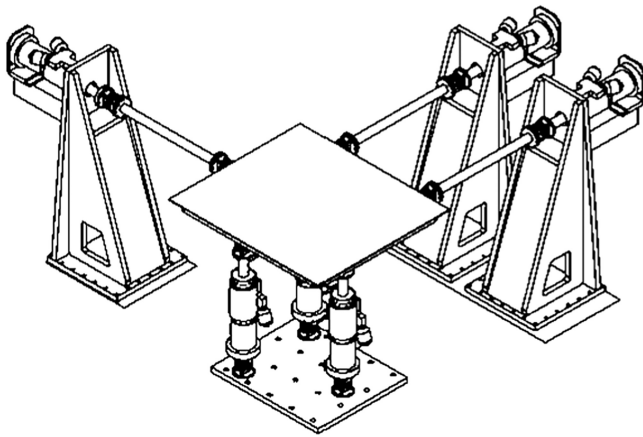


FIGURE 1—Schematic illustration of the MAST 9720.

was developed, which comprised various acceleration conditions (Table 1) to replicate a range of physiological movements. These were applied to the units by manipulating the frequency of oscillation and displacement amplitudes. Most human movements tend to fall between 0.3 and 3.5 Hz (34), and maximum angular velocities of the forearm during the drive phase in elite wheelchair racers have a frequency component of 3.6 Hz (37). The conditions selected produced similar PAC to those recorded by a GT3X+ device worn on the wrist during wheelchair propulsion, ranging from light- to moderate-intensity PA based on MET values. In some conditions, acceleration was kept constant, allowing for an independent assessment of the effect of frequency on count magnitude. Conversely, frequency was also kept constant while acceleration of the aluminum mounting plate was manipulated to independently assess the effect of acceleration on accelerometer output. The units were subjected to these testing conditions using a sinusoidal oscillation procedure, which was preprogrammed using the MAST-dedicated software (RS Replay; Instron Structural Testing Systems Ltd.).

All 10 GT3X+ units to be used in subsequent human testing were initialized at a sampling frequency of 30 Hz via a computer interface. Before testing, a trial run was conducted without the accelerometers to ensure that the hydraulics were functioning at an optimum operating temperature. Then, each

GT3X+ unit was attached to a piece of angle iron 2 cm apart using a double-sided floor tape (DS Scrim 306/250; Tape Range Distributors Ltd., Cheshire, United Kingdom). Each GT3X+ unit was mounted firmly and securely to prevent accelerometer misalignment. The angle iron was then attached to the aluminum alloy mounting plate of the MAST. Because only the horizontal actuator was used during this protocol, the angle iron was rotated after each test schedule so that the horizontal motion of the MAST corresponded to displacement along the X, Y, and Z axes of the GT3X+. The testing schedule was repeated in all three axes. Each individual condition was maintained for 70 s. After completion of the testing schedule and when the MAST rig was safely parked, the accelerometers were removed and downloaded using a dedicated software (Actilife 6; Pensacola, FL) and exported to Microsoft Excel in a comma-separated value file format for further analysis. The activity counts were summated into 5-s epochs. The first and last 10 s were excluded to ensure that only steady-state values were included in the analysis. Each condition was reduced to 50 s, with the mean of the 10 remaining values expressed as counts per 5 s.

Experiment 2—Human Validity

Ethics approval was granted by the University of Bath Research Ethics Approval Committee for Health, and an informed consent was obtained from each participant. Fifteen MWU (mean $\pm$ SD: age, 36 ± 11 yr; time since injury, 15 ± 17 yr; body mass, 70 ± 12 kg) visited the Centre for Disability Sport and Health human physiology laboratory on one morning after an overnight fast. The medical condition responsible for regular use of a wheelchair was nine SCI (paraplegic), one fibromyalgia, one complex regional pain syndrome, and two participants with spina bifida. Two able-bodied participants were included in the analysis; both were familiar with wheelchair propulsion, having played wheelchair basketball for >1 yr. Time since injury was self-reported on the basis of the time when the medical condition was first diagnosed by a clinician. Anthropometric variables were collected, and resting metabolic rate (RMR) was measured in a semirecumbent position in accordance with best practice (9).

TABLE 1. Description of the acceleration and frequency conditions used during the mechanical testing schedule and within-trial intra- and interunit CV values.

Condition	Amplitude (m)	Frequency (Hz)	Acceleration (g)	Intraunit CV			Interunit CV		
				Axis			Axis		
				Z	Y	X	Z	Y	X
1	0.063	0.5	0.06	4.4% (3.4–5.4)	2.5% (2.1–3.0)	4.7% (3.5–5.9)	5.2% (4.4–6.0)	4.7% (4.1–5.4)	5.0% (4.0–6.0)
2	0.063	1.0	0.25	0.9% (0.7–1.1)	0.5% (0.3–0.7)	0.8% (0.7–1.0)	1.6% (1.3–1.9)	0.9% (0.8–1.0)	1.1% (0.9–1.3)
3	0.055	1.5	0.50	0.4% (0.3–0.4)	0.3% (0.3–0.4)	0.5% (0.3–0.7)	1.8% (1.7–2.0)	1.9% (1.7–2.0)	3.3% (3.0–3.5)
4	0.031	2.0	0.50	0.5% (0.3–0.6)	0.6% (0.3–0.8)	0.4% (0.4–0.5)	1.4% (1.3–1.5)	1.2% (1.0–1.4)	1.8% (1.7–1.9)
5	0.062	2.0	1.00	0.2% (0.2–0.3)	0.2% (0.1–0.3)	0.2% (0.2–0.3)	1.1% (1.0–1.1)	1.4% (1.3–1.5)	1.0% (1.0–1.1)
6	0.040	2.5	1.00	0.3% (0.2–0.4)	0.3% (0.1–0.5)	0.4% (0.2–0.5)	2.5% (2.3–2.6)	2.7% (2.6–2.8)	4.1% (3.9–4.2)
7	0.016	4.0	1.00	1.6% (1.2–2.1)	1.5% (1.1–1.8)	1.5% (1.1–2.0)	2.9% (2.6–3.2)	5.3% (5.0–5.6)	3.6% (3.2–4.0)
8	0.035	3.0	1.25	0.5% (0.4–0.7)	0.3% (0.2–0.5)	0.3% (0.2–0.5)	3.9% (3.7–4.1)	3.5% (3.5–3.6)	3.4% (3.2–3.5)
9	0.025	3.5	1.25	0.7% (0.6–0.8)	0.6% (0.4–0.8)	0.6% (0.5–0.7)	1.0% (0.8–1.2)	1.7% (1.7–1.8)	1.4% (1.3–1.6)
10	0.060	2.5	1.50	0.2% (0.2–0.3)	0.3% (0.2–0.3)	0.3% (0.2–0.3)	1.7% (1.7–1.8)	0.9% (0.8–0.9)	1.8% (1.7–1.8)
11	0.023	4.0	1.50	0.7% (0.5–0.9)	0.7% (0.6–0.8)	0.8% (0.6–1.0)	2.0% (1.9–2.2)	2.3% (2.2–2.5)	2.8% (2.6–2.9)
Overall mean				0.9% (0.7–1.2)	0.7% (0.5–0.9)	1.0% (0.7–1.2)	2.3% (2.1–2.5)	2.4% (2.2–2.6)	2.7% (2.4–2.9)

Data are displayed as mean and 95% confidence intervals (lower to upper).

Activity Protocol

The activity protocol consisted of wheelchair propulsion at varying velocities counterclockwise around an outdoor athletics track and simulated desk work. This created a controlled research environment but, importantly, outside the laboratory, where energy expenditure is more likely to reflect that of daily wheelchair propulsion. Each activity lasted 6 min, interspersed with 5-min recovery periods. Throughout the activity protocol, three GT3X+ units were worn, one on the right wrist (using a Velcro wrist strap positioned over the dorsal aspect of the wrist midway between the radial and ulnar styloid processes), one on the upper arm (using a small elastic belt positioned on the lateral surface of the arm midway between the acromion process and lateral epicondyle of the humerus), and one on the waist (positioned above the right hip along the anterior axillary line). The devices were initialized with a sampling frequency of 30 Hz.

In addition, participants also wore a portable metabolic system (COSMED K4b²; COSMED, Rome, Italy) and a Polar® Team HR monitor (Polar Electro, Inc., Lake Success, NY). A rubber face mask (Hans Rudolph, Inc., Shawnee, KS) of appropriate size was fitted carefully to the face and checked for leaks before each test. Expired gases pass through a flow meter and are channeled down a Perma Pure sampling line into the analyzer unit where the fractions of O₂ and CO₂ in expired air are measured. Metabolic data were retrieved and analyzed using an associated software (COSMED 9.0; COSMED, Rome, Italy). Oxygen uptake ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) were used to estimate energy expenditure (kJ·min⁻¹) of each activity using indirect calorimetry (40). Before use, the K4b² was calibrated according to manufacturer's instructions.

The wheelchair propulsion activities included four conditions, 2, 4, 6, and 8 km·h⁻¹, which were counterbalanced to prevent order and carry-over effects using a Latin square design. During desk work, participants were asked to type out a script. Participants only completed trials that they felt comfortable/competent with. Real-time speed feedback was provided via a GPS cycle computer (Garmin® EDGE 500; Garmin Ltd., Southampton, United Kingdom) placed where visible in the participants' lap. No attempt was made to standardize wheelchair variables, although tire pressure and chair characteristics were recorded, and participants used their everyday wheelchair. As alluded to elsewhere (39), differences in these variables, such as chair weight, would be reflected in oxygen uptake values.

Statistical Analyses

Experiment 1—mechanical testing. The mean $\pm$ SD activity PAC output was calculated for each unit in each condition and each axis (330 in total). The coefficient of variation (CV_{intra}) was calculated from the replicate 5-s epochs within each condition to assess intraunit reliability. This is a noteworthy distinction of our design compared with those of previous research in the field of intraunit reliability

analyses, which tend to focus on within-unit, between-trial variability (18). However, we adopted an approach similar to that of Esliger and Tremblay (11) to remove any trial effects that may increase variability (i.e., more technological error). Second, CV for each axis (CV_{inter}) during each condition were determined. In addition, intraclass correlation coefficients (ICC) with a two-way random effects model for absolute agreement were calculated.

A Spearman rank correlation coefficient (R_s) was used to determine the criterion-related validity between PAC from the GT3X+ and the MAST acceleration. Paired *t*-tests were conducted to assess the independent effect of acceleration when frequency was held constant at 2 Hz on PAC output across units. A repeated-measures ANOVA was conducted to assess the independent effect of frequency on PAC output across units when acceleration was held constant at 9.81 m·s⁻². Where significance was found ($P < 0.05$), Bonferroni corrections were applied to *post hoc* tests where multiple comparisons were considered.

Experiment 2—human validity. *A priori* power calculation revealed that a sample size of 15 was necessary to detect an *r* of 0.67 using a one-tailed test with $\alpha = 0.05$ and power = 0.95. This calculation was based on data from a previous study (39). The K4b² and activity monitors were synchronized before use. Breath-by-breath K4b² data were interpolated into 1-s intervals for all tests. Individual $\dot{V}O_2$ and $\dot{V}CO_2$ breath values that were >3 SD from the mean were removed (22). Final data sets were then averaged over a 2-min period. PAC from the GT3X+ were summated into 60-s epochs. Assuming that diet-induced thermogenesis was negligible (i.e., participants were fasted), RMR (kJ·min⁻¹) was subtracted from total energy expenditure (TEE) measured by the K4b² to generate PAEE for each activity. Comparisons between the “criterion” measurement of PAEE (TEE – RMR) and activity monitors were made between 03:30 and 05:30 (mm:ss) of each activity.

Pearson product moment correlation coefficients (*r*), coefficients of determination (R^2), and linear regressions were conducted to assess the association between the criterion and PAC from the GT3X+ accelerometers at each anatomical position during wheelchair propulsion. Using the generated regression equations, an analysis of agreement was conducted for each anatomical location using Bland–Altman plots to calculate absolute bias and 95% limits of agreement (LoA). SEE was also calculated for each correlation. Statistical significance was set *a priori* at $\alpha < 0.05$. All analyses were performed using IBM® SPSS® Statistics 20 for Windows (IBM, Armonk, NY).

RESULTS

Experiment 1—mechanical testing. The overall mean $\pm$ SD activity counts across all 11 testing conditions for all devices were 497 ± 2.4 , 497 ± 2.0 , and 496 ± 2.4 counts per 5 s for the Z, Y, and X axes, respectively. Intraunit reliability (CV_{intra}) values, displayed as mean and 95% confidence

intervals (lower to upper), were 0.9% (0.7–1.2), 0.7% (0.5–0.9), and 1.0% (0.7–1.2) for the *Z*, *Y*, and *X* axes, respectively (Table 1). Irrespective of the axis, the highest and lowest CV_{intra} values corresponded to conditions 1 (0.06g, 0.5 Hz) and 5 (1.0g, 2.0 Hz), respectively. We also considered the between-trial intraunit reliability values, which were higher than within-trial results. These were 1.5% (0.8–2.2), 1.5% (0.7–3.1), and 1.7% (0.8–2.5) for the *Z*, *Y*, and *X* axes, respectively (mean, 95% upper and lower confidence intervals).

The ICC for activity counts across all conditions was 1.0 for each axis (all $P < 0.001$). The mean variability between units was 2.5% (CV_{inter}) across all conditions for all units in each axis and ranged from 1.0% to 5.2%, 0.9% to 5.3%, and 1.0% to 5.0% for the *Z*, *Y*, and *X* axes, respectively (Table 2).

Figure 2 demonstrates a significant, weak, positive linear relation ($R_s = 0.25$, $P < 0.01$) when PAC outputs across all 11 conditions from each axis for all units are displayed together ($n = 660$). Holding the frequency of oscillation of the mounting plate of the MAST constant at 2 Hz and increasing acceleration lead to a significant increase in PAC ($0.5g = 462 \pm 2$ counts per 5 s, $1.0g = 977 \pm 2$ counts per 5 s, $P < 0.01$). However, holding acceleration constant at $9.81 \text{ m}\cdot\text{s}^{-2}$ and manipulating the frequency of movement had counterintuitive results; interestingly, increasing movement frequency resulted in a significant decrease in PAC (2 Hz = 977 ± 2 counts per 5 s, 2.5 Hz = 644 ± 2 counts per 5 s, 4 Hz = 147 ± 2 counts per 5 s, $P < 0.01$).

Experiment 2—human validity. The relations between criterion PAEE estimated by the K4b² and predicted PAEE derived from activity counts from each device are presented as scatter plots in Figure 3A–C. PAC from each anatomical location were significantly ($P < 0.01$) associated with PAEE (waist, $r = 0.73$; upper arm, $r = 0.87$; wrist, $r = 0.93$). This relation remains linear when assessing wheelchair propulsion separately (wrist, $r = 0.90$). The SEE for each correlation were 6.07, 4.38, and $3.34 \text{ kJ}\cdot\text{min}^{-1}$ for the waist, upper arm, and wrist, respectively. The linear regression equations for devices worn at each anatomical location are shown in equations 1–3.

$$PAEE_{\text{waist}} = (0.004815 \times \text{PAC}\cdot\text{min}^{-1}) + 5.294092 \quad [1]$$

$$PAEE_{\text{upper arm}} = (0.001642 \times \text{PAC}\cdot\text{min}^{-1}) + 0.204579 \quad [2]$$

$$PAEE_{\text{wrist}} = (0.000929 \times \text{PAC}\cdot\text{min}^{-1}) - 0.284818 \quad [3]$$

Figure 4A–C further illustrates the difference between the criterion PAEE and the predicted PAEE through the use of

Bland–Altman plots displaying the mean difference and 95% LoA. Using the generated regression equations, the absolute bias $\pm 95\%$ LoA values were $0.0 \pm 11.8 \text{ kJ}\cdot\text{min}^{-1}$, $0.0 \pm 8.5 \text{ kJ}\cdot\text{min}^{-1}$, and $0.0 \pm 6.5 \text{ kJ}\cdot\text{min}^{-1}$ for the waist, upper arm, and wrist, respectively.

DISCUSSION

Of the three anatomical locations considered in this study, the results indicate that the wrist provides the most valid prediction of PAEE in MWU. The accelerometer worn on the wrist explained the highest amount of variance and displayed the lowest random error. Using a schedule that comprised 11 test conditions of various frequencies and accelerations, the GT3X+ demonstrated excellent reliability, with mean intra- and interunit CV of 0.9% and 2.5%, respectively. To our knowledge, this is the first study to assess the mechanical reliability and validity of the newest-generation ActiGraph™ GT3X+ accelerometer and validity of its use in MWU.

Mechanical testing. The majority of previous mechanical reliability studies have focused on older generations of the ActiGraph accelerometer, of which the GT1M displays the next best reliability compared with that in the GT3X+ ($CV_{\text{intra}} = 2.9\%$ and $CV_{\text{inter}} = 3.5\%$) (31). Considering the aforementioned software and component improvements, such as switching to microelectromechanical system transducers, which have greater sensitivity, it is not surprising that newer generations display improved intra- and interunit reliability. The older generations of the ActiGraph accelerometers contained piezoelectric transducers, which were typically fitted manually during manufacturing by experienced technicians (7). Intuitively, this might explain the increased interunit variability with older models. Older generations (7164 model) have also demonstrated large interunit variation ($>100\%$) at lower accelerations ($<1 \text{ m}\cdot\text{s}^{-2}$) (5). Although not of the same magnitude, our results also indicate poorer interunit reliability during the lowest frequency and acceleration condition (5%). However, a recent study assessing the interunit reliability of the GT3X model reported a mean CV_{inter} of 60.2% across a range of accelerations and was $>149.4\%$ when units were oscillated at 1.1 Hz (29). This is disconcerting, especially considering that the improvements with the newer-generation GT3X+ are mostly cosmetic. We have displayed acceptable interunit reliability; only three out of the 33 conditions tested displayed $CV_{\text{inter}} \geq 5\%$. These variances could be explained by differences in the protocol, whereby Santos-Lozano et al. (29) included a

TABLE 2. GPS velocity, criterion PAEE, PAC at each anatomical location, calculated METs, HR, and number of participants per trial for each activity (mean $\pm$ SD).

Activity	Garmin Velocity ($\text{km}\cdot\text{h}^{-1}$)	K4b ² PAEE ($\text{kJ}\cdot\text{min}^{-1}$)	PAC (counts $\cdot\text{min}^{-1}$)			MET K4b ²	HR (bpm)	<i>n</i>
			Upper Arm	Waist	Wrist			
Desk work		1.1 ± 1.1	30 ± 38	1 ± 3	362 ± 182	1.3 ± 0.3	73 ± 14	15
2 $\text{km}\cdot\text{h}^{-1}$	2.6 ± 0.4	5.3 ± 1.7	5748 ± 1399	644 ± 757	8192 ± 2209	2.3 ± 0.4	96 ± 19	12
4 $\text{km}\cdot\text{h}^{-1}$	4.0 ± 0.3	10.1 ± 3.9	7098 ± 2168	841 ± 643	$11,712 \pm 3313$	3.5 ± 1.1	105 ± 17	15
6 $\text{km}\cdot\text{h}^{-1}$	5.6 ± 0.6	15.7 ± 4.2	8477 ± 2054	1803 ± 1347	$17,105 \pm 4271$	4.7 ± 1.2	125 ± 23	13
8 $\text{km}\cdot\text{h}^{-1}$	7.1 ± 0.7	24.8 ± 7.9	$12,459 \pm 5042$	2880 ± 1421	$25,599 \pm 4522$	6.4 ± 2.1	149 ± 22	9

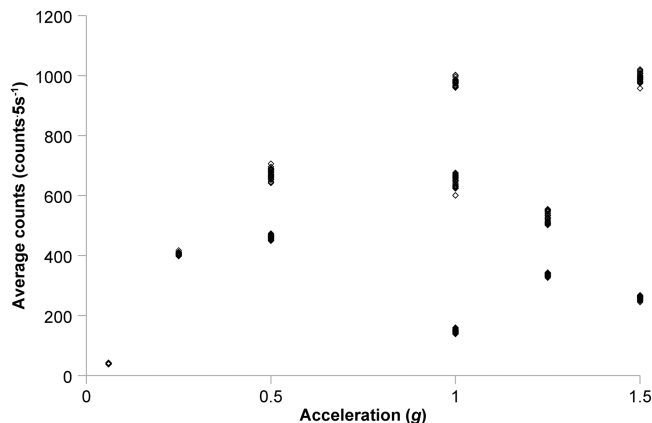


FIGURE 2—Relation between count magnitude and MAST rig acceleration ($n = 660$).

condition outside the range of human motion (10 Hz) contributing to the higher overall mean CV_{inter} . The authors also used a smaller-scale vibration table as opposed to a MAST rig; it is unclear whether the vibration table was separated from the electric motor to minimize the mechanical vibration, as advocated previously (10). Moreover, we believe that there is an error in the calculation of accelerations used in the Santos-Lozano et al. (29) study. They cite a 1.1-Hz site of orbit and a 0.04-m radius of orbit and claim that this yields an acceleration of 1.087g. Using the standard equation for

tangential acceleration (equation 4), we calculate the acceleration for this condition as 0.194g.

$$u = 4\pi^2/2r \quad [4]$$

In our study, the ICC observed across testing conditions were high and concurred with those reported for other available accelerometers (18,26) and previous generations of ActiGraph (5). If interunit reliability is poor, then it becomes difficult to distinguish whether the variability in PA during free-living monitoring between subjects is solely attributed to variations in behavior or inherent to the accelerometer. The GT3X+ demonstrated poor validity when compared with that of criterion acceleration of the MAST rig (Fig. 2). The weak relation of $R_s = 0.25$ between 0.06g and 1.5g is well below that of most industry-standard PA monitoring accelerometers. The GENE and 3DNX PAC outputs are both strongly related to acceleration during a mechanical setup, with correlations of $r = 0.97$ and $r = 0.99$, respectively (10,18). In the older 7164 model ActiGraph, Brage et al. (5) unequivocally stated that count output is only proportional to acceleration if frequency is held constant, implying that some form of frequency-dependent filter exists. This would also seem to hold true for the newest-generation GT3X+, supported by our counterintuitive findings of decreased PAC as frequency of oscillation is increased while acceleration was held constant. Brage et al. (5) developed and then used a frequency-based correction factor, which, when applied to ActiGraph counts,

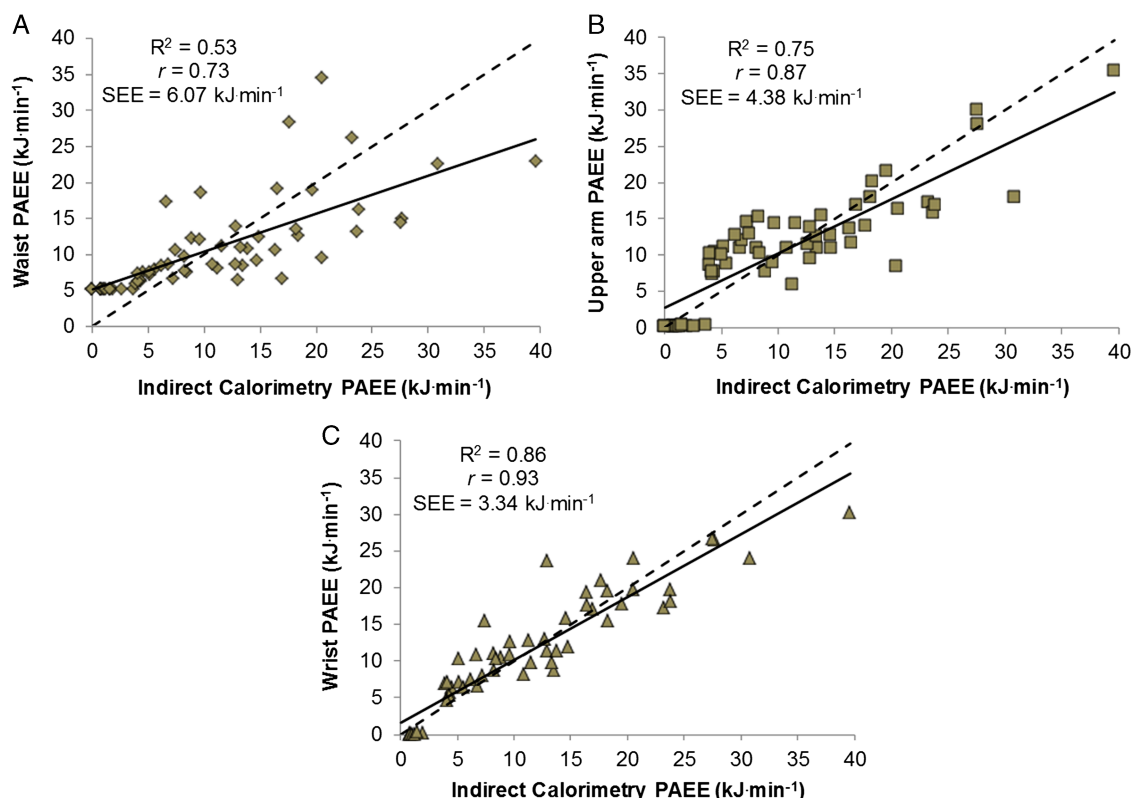


FIGURE 3—Scatter plots showing the relation between predicted PAEE from the waist (A), upper arm (B), and wrist (C) against the criterion PAEE. The *straight line* represents the models best fit, and the *dotted line* indicates the line of identity.

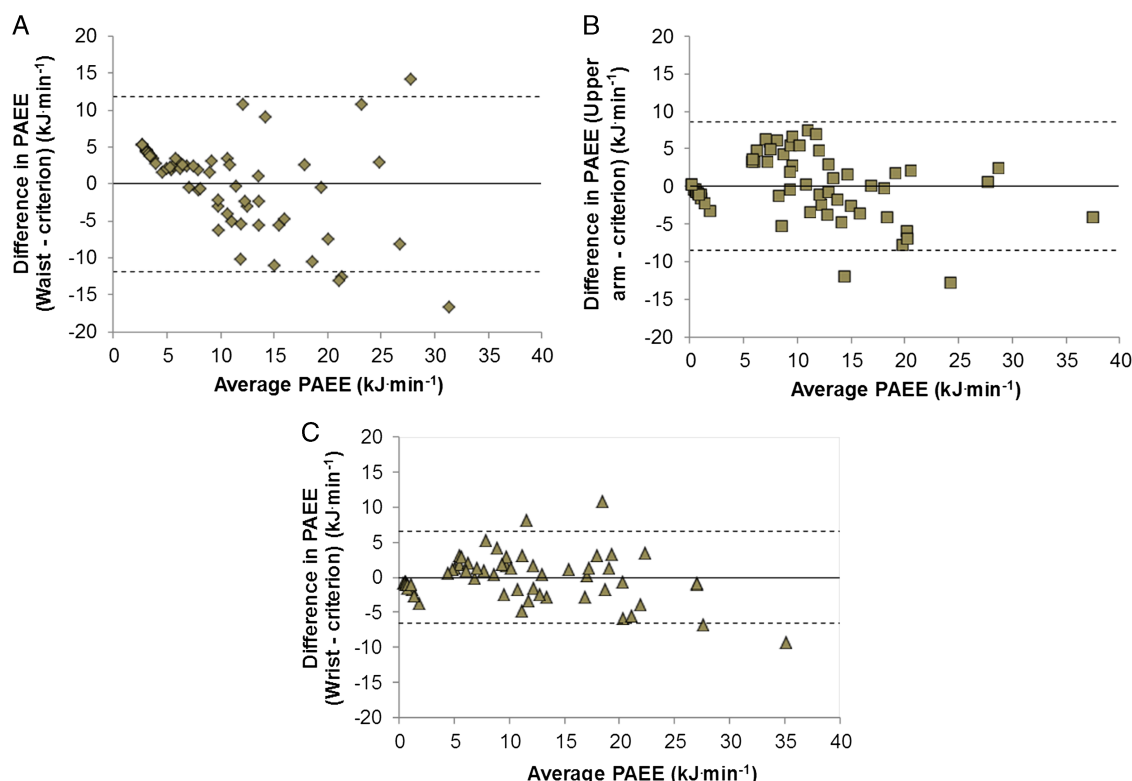


FIGURE 4—Bland–Altman plots for the criterion and estimated PAEE using regression equations developed at the waist (A), upper arm (B), and wrist (C).

restores linearity, improving the relation from $r = 0.69$ to $r = 0.94$. Most of the current accelerometers on the market use band-pass filters to extract acceleration signals within certain frequency ranges while discarding those that are not likely to be representative of “human movement.” Outputs from other accelerometers including the Actical (11) and RT3 (26) have also been shown to be dependent on movement frequency in a mechanical setup.

Experiment 2—human validity. Hiremath and Ding (16) advocated the importance of keeping the accelerometer device constant but manipulating its anatomical positioning to determine the most appropriate placement of an accelerometer to capture PAEE in MWU. Our results indicate that of the three anatomical locations considered, a wrist-mounted accelerometer provides the most valid prediction of PAEE during outdoor wheelchair propulsion. This is the first study to assess the validity of the GT3X+ accelerometer in this population and to evaluate the accuracy of specifically developed algorithms capable of predicting PAEE.

Accurate measurement of habitual PAEE is a prerequisite to determine the link and establish dose–response relations between PA and health (33). Surprisingly, relatively few studies have tried to evaluate monitoring tools among wheelchair users (38,39). Washburn and Copay (39) found that PAC from the older-generation uniaxial CSA accelerometer worn on the wrist had a moderate relation (left wrist, $r = 0.66$; right wrist, $r = 0.52$; $P < 0.01$) with oxygen uptake during wheelchair propulsion at three velocities. Warms and

Belza (38) observed low-to-moderate relations ($r = 0.30$ – 0.77 , $P < 0.01$) between activity counts from an Actiwatch containing an omnidirectional accelerometer and self-reported PA. Although these results suggest that an accelerometer located on the wrist is a suitable measure of PAEE for individuals with an SCI, the study of Warms and Belza (38) is only able to confer the concurrent validity of wrist actigraphy against a self-reported measure of activity intensity and frequency. The higher correlation ($r = 0.93$) observed between activity counts at the wrist and PAEE in this current study might be due to the direct comparison against a criterion measurement of PAEE rather than a self-reported measurement or the inclusion of more than three propulsion velocities and an activity in daily living of low intensity, combined with using a triaxial accelerometer offering greater sensitivity.

A similar relation was observed at the upper arm ($r = 0.87$) compared with that in previous research using an RT3 triaxial accelerometer ($r = 0.83$) (17). It is perhaps pertinent to address some methodological differences here because the authors compared activity counts against TEE (including RMR and diet-induced thermogenesis) and not PAEE as measured in the present study. TEE and PAEE should not be equated (35). It is a noteworthy distinction that accelerometers are only capable of detecting movement and should therefore be associated only with the component of energy expenditure arising from skeletal muscle contraction-induced movements.

Furthermore, PA monitoring devices (e.g., Actiheart) use demographic characteristics such as body mass to predict RMR and determine TEE. Common equations to predict RMR in the general population are inappropriate to use for individuals with an SCI and have been shown to overpredict measured requirements by 5%–32% (6). Considering that RMR is the largest component of TEE particularly in sedentary individuals (36), the error observed with the prediction of TEE by these devices in MWU may be a result of the algorithms used to determine RMR not being suitable for individuals with an SCI.

Furthermore, although the previous studies made no attempt to control for individual variations in RMR, they also only reported correlations and made no attempt to develop regression equations capable of accurately predicting PAEE. This study attempted to build on this by assessing the degree of error associated with the generated equations for PAC at each anatomical location. The mean bias for each location was negligible. However, these findings should be viewed with caution. Because the development of the regression equations to predict PAEE and subsequent evaluation were carried out on the same sample of participants, there is a tendency for the evaluation statistics to be biased and be overly optimistic (32). Therefore, we appreciate that further work is required to cross-validate these equations on an independent sample of wheelchair users.

However, considering that the primary aim of this study was to identify the best anatomical location to capture PAEE, analyzing our results this way can offer an insight into the spread of random error. For example, visual inspection of Figure 4A indicates a considerable degree of heteroscedasticity using the prediction equation at the waist. Thus, this anatomical location displayed increased random error as the intensity of activity increased. Despite the wrist displaying the narrowest 95% LoA (Fig. 4C) ($\pm 6.5 \text{ kJ}\cdot\text{min}^{-1}$), it is advisable, when more studies have been published in the area, that the academic community produce a consensus statement addressing the clinical limits of PA assessment in this population. However, in combination with the highest association to the criterion measurement and lowest SEE reported, these data suggest that the wrist is the most appropriate anatomical location to quantify PAEE in MWU. With movement restricted to the upper limbs in MWU, the most distal anatomical location seemingly offers improved sensitivity to the detection of PAEE during wheelchair propulsion.

ActiGraph PAC have been shown to peak at approximately $10\text{--}12 \text{ km}\cdot\text{h}^{-1}$ when running and plateau thereafter when worn at the hip in ambulatory subjects (20). Knowledge regarding digital signal processing filters has only recently become more available because an obligation has been placed on device manufacturers to be more transparent with regard to their specific properties and functions. The GT3X+ has half-power frequencies of 0.5 and 2.5 Hz; taken from the device manual, it could be misleading that movements within these limits are measured in full scale whereas those outside it are not registered at all. Larger bandwidth

filters could allow physiologically unrelated vibrations or noise to be included in the signal. Conversely, overly aggressive, frequency-dependent filtering can lead to erroneous measurements of human movements and cause the previously observed plateau effect (27). However, a plateau effect was not observed when worn at the wrist for speeds up to $8 \text{ km}\cdot\text{h}^{-1}$ yet the study cannot conclude whether a plateau phenomenon exists above this propulsion speed. Only nine of 15 participants (two of which were able bodied) were able to complete the propulsion speed of $8 \text{ km}\cdot\text{h}^{-1}$. Considering that wheelchair users have been shown to achieve minimal amounts of strenuous activity during free living (38), if a plateau effect does exist, it may negligibly affect the accuracy of monitoring PAEE in this population.

A limitation of this study is the relatively small sample size and considerable variation within subjects based on the diversity of disabilities included. However, this diversity may be considered beneficial because the assortment of the propulsion techniques captured improves the external validity of the regression equations, making them more suitable for the wider wheelchair user population. Once we removed the two occasional wheelchair users from the analysis, this had a negligible effect on the relations observed between PAEE and PAC at the wrist (all data, $r = 0.93$; regular MWU only, $r = 0.92$); consequently, we see value in taking a more generic approach. Also, despite the diversity of the population, the amount of unexplained random error is relatively small. The inclusion of a diverse range of subjects is in accordance with best practice recommendations for PA validation studies (1). Future studies should assess the validity of the GT3X+ for predicting PAEE during more complex representative daily activities performed by MWU to determine whether band-pass filtering processes may affect the sensitivity of the GT3X+ to quantify sedentary behaviors or detect vigorous-intensity activities above a certain threshold. The devices' use to accurately assess PAEE during free living also needs to be explored. The associated equations, which are generated, require cross-validation using an independent sample of representative participants. Furthermore, mechanical testing was conducted over a limited range of accelerations ($0.06\text{g}\text{--}1.5\text{g}$) using simple single-axis movements, which do not cover the entire dynamic range of the GT3X+ device ($\pm 6\text{g}$) or the complete range within which physiologically relevant movements can occur (3). ActiGraph has assured us that, on the basis of comprehensive testing during manufacturing, their GT3X+ devices are stable over time. Future studies should undertake a more comprehensive testing schedule across the devices' entire dynamic range for longer durations (e.g., $>6 \text{ h}$) and potentially during more complex three-dimensional movements to determine simulated performance over longer durations.

In conclusion, we have shown excellent intra- and inter-unit reliability of the GT3X+. Although the unidimensional mechanical test data are useful in evaluating the devices' mechanical reliability, it is important to remember that the

GT3X+ is an accelerometry-based PA monitor. Our applied data from human testing suggest that it is a valid tool for predicting PAEE during complex multidimensional human movements, such as wheelchair propulsion. Of the three anatomical locations considered, a wrist-mounted accelerometer provides the most accurate prediction of PAEE in MWU during propulsion.

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