A new wheelchair ergometer called the Wheelchair Aerobic Fitness Trainer (WAFT) has been developed. Commercially available magnetic eddy current braking devices (Mi-

noura MAGTURBO™) were incorporated into the wheelchair ergometer as a cost-effective and reliable method of creating variable power output for graded exercise testing and aerobic conditioning. Modifications to the standard MAGTURBO™ were necessary in order to simulate the physical stress normally experienced during daily wheelchair propulsion. The WAFT with modified MAGTURBO™ offers independent manipulation of speed and resistance for each wheel. This flexibility makes the WAFT appropriate for graded exercise testing and conditioning of persons with lower limb disabilities who have widely varied exercise capacities.

This paper describes the WAFT and a series of calibration experiments designed to identify the operating characteristics of the device and to establish regression equations for calculating power output as a function of wheel speed and MAGTURBO™ resistance setting. Based on these power output relations, graded wheelchair exercise test protocols were defined for the WAFT, and results from tests performed by persons with a variety of lower limb disabilities and a wide range of cardiorespiratory fitness are reported.

Index Terms—wheelchair ergometry, spinal cord injury, exercise testing.

I. INTRODUCTION

A VARIETY of equipment is available for the assessment of aerobic capacity and cardiovascular health of the able-bodied, including computer-controlled treadmills and mechanically and electrically braked cycle ergometers. These devices, coupled with standardized testing protocols, have found widespread use and acceptance. However, there is a shortage of appropriate testing equipment for determining the cardiorespiratory fitness of persons limited to upper body exercise because of disabilities resulting from lower limb amputation, musculo-skeletal and neurological impairments, peripheral vascular disease, and spinal cord injury. It is well known that the spinal cord injured wheelchair user and other lower limb disabled are frequently sedentary. Lack of physical activity leads to significant decrements in physical fitness and further increases the risk of cardiovascular disease [1]–[4].

Exercise testing protocols have not been standardized for the lower limb disabled population; testing equipment is costly, cumbersome, and difficult or impossible to calibrate; and testing procedures frequently employ a mode of exercise that is not specific to the physiological demands or biomechanics of wheelchair propulsion. Moreover, equipment that would satisfy the requirements for upper body exercise testing has been limited to use in research studies and is not available for daily clinical diagnostic testing or for use in patient rehabilitation programs.

We have conducted a series of projects to (a) design, construct, and calibrate a wheelchair ergometer called the Wheelchair Aerobic Fitness Trainer (WAFT) and (b) develop standardized maximal wheelchair ergometer exercise testing protocols. The purpose of this paper is to describe the wheelchair ergometer calibration procedures and report the results of a series of speed-torque calibration experiments for the magnetic eddy current braking assembly and wheelchair ergometer. These calibration experiments were used to establish regression equations for calculating the ergometer power outputs at various combinations of speed and resistance settings. Knowledge of the power outputs achieved by persons with lower limb disabilities during wheelchair graded exercise testing is useful in (a) developing exercise prescriptions, (b) evaluating the effectiveness of conditioning or rehabilitation programs, (c) charting the effects of disability on functional capacity, (d) formulating a procedure for predicting peak exercise capacity from submaximal wheelchair graded exercise testing, and (e) conducting exercise physiology research. Two graded exercise test protocols for the WAFT were developed from power output data obtained from the calibration experiments described herein, and comparison of arm crank versus WAFT exercise testing using these protocols is discussed briefly.

II. EQUIPMENT DESCRIPTION

A. Description of the Wheelchair Ergometer

The wheelchair ergometer (Fig. 1(a)) was designed and constructed at VA Hines Hospital, Rehabilitation R&D Center (Hines, IL). Improvements in mechanical and electronic design have resulted in several generations of this device since it was first reported [5]. The loading ramps of the WAFT can be
adjusted for different sizes of standard, lightweight, and sport wheelchairs. The ramps are tilted during wheelchair loading and unloading. The wheelchair is backed up the ramps until the rear wheels rest on three rollers built into each ramp. The three rollers have been configured to accept a 61-cm-diameter wheel and provide an even distribution of the weight of chair and rider during wheeling. When the ramps are in the down position, the indentations used to position the front casters are closed and the forward-most roller is locked so it cannot turn (Fig. 1(b)). Two levers, one on each ramp, are pushed forward to level the ramps and form a stable platform. In the level position, the rear wheelchair wheels turn freely on the rollers and the front wheels are secured in the open indentations (Fig. 1(c)).

Each rear wheel of the wheelchair is cradled and rotates on three rollers, 16 cm long by 10 cm in diameter. A three roller configuration was adopted in order to reduce the problematic rolling friction present in two-roller systems. Additionally, balanced weights were placed inside each of the six rollers in an effort to replicate the translational inertia that normally occurs during wheelchair operation.

Fig. 1. Schematic of the Wheelchair Aerobic Fitness Trainer (WAFT). (a) Device in loading position with all components: 1) magnetic eddy current brake, 2) brake controls and levers for raising and lowering ramps, 3) electronic speedometer for right and left wheels, 4) linkage between ramps, 5) front wheel anchor plate, 6) loading ramps, 7) anchor bar for lateral adjustment of ramps, 8) front support legs for ramps, and 9) non-skid base mat. (b), (c) Cutaway view showing configuration of inertially loaded rollers, and linkage for raising and lowering ramps and positioning front support legs (11). (d), (e) Modifications made in the MAGTURBOTM magnetic eddy current brake: timing belt (12) and tension roller (10) disengaged for freewheeling (d); timing belt (12) and tension roller (10) engaged (e). (f) Side view of one roller with reflective opto-interrupters (13) positioned for the detection of wheel speed and direction.

B. Magnetic Eddy Current Braking

Various techniques have been employed to create resistance in cycle ergometers: frictional drag of a belt held against a flywheel, electrical braking, fan assemblies for air braking, and hydraulic braking. On the WAFT, braking resistance for each rear wheelchair wheel is created through the use of a MAGTURBOTM magnetic eddy current braking assembly. Within each MAGTURBOTM unit, two arrays of equally spaced magnets are positioned on each side of an aluminum disk. One array of magnets can be moved in 10° increments while the other remains in a fixed position. The aluminum disk is mounted on a small axle that is coupled to the rear-most roller of each ramp via a timing belt (see Fig. 1(d), 1(e), and Calibration Results). When the wheelchair wheels are turned, the rollers rotate and the aluminum disk spins in the variable magnetic field. Faraday's law of electromagnetic induction and Lorentz's force law explain the magnetic drag force used to create variable resistance in devices similar to the MAGTURBOTM [6]. Controls were installed to disengage the drive belt, so that a wheelchair could be "free-wheeled" on the WAFT (Fig. 1(d) and 1(e)). This form of braking was selected since it provides a trouble free, user safe, relatively inexpensive and uncomplicated system.

The MAGTURBOTM has seven user-controlled resistance settings. At each setting the resistance created by the MAGTURBOTM varies monotonically with the velocity of the wheelchair wheels. An inertial flywheel is factory mounted on the axle of each MAGTURBOTM. Based on the qualitative judgments of a number of manual wheelchair users who tested the configuration, the inertial flywheel, weighted rollers, and magnetic eddy current brakes appear to produce conditions that simulate actual manual wheelchair propulsion. The resistance and speed of each wheel may be manipulated independently to accommodate the user's objectives and physical abilities.

C. Electronics and Computer Interface

Two reflective opto-interrupters are mounted next to the rollers on either side of the WAFT (Fig. 1(f)). The opto-interrupters sense the presence of eight timing marks equally spaced around the center roller. The pulse trains generated by these opto-interrupters are input to a frequency-to-voltage converter (National 2907), and a voltage proportional to wheel speed \( \nu = 1 \) is produced for each wheel (Fig. 2). This voltage is used to drive a small panel meter that serves as a speedometer for feedback to the user. Each wheel has its own speedometer; the two are housed in the same enclosure for ease in viewing (Fig. 1(a)).

A PC Limited 200 computer was interfaced with the WAFT via a Metabyte DASH-16 A/D Board to sample the output from the voltage-to-frequency converter. Programs written in ASYST, a scientific programming language, sample the wheel speed voltages from each side every 0.2 s [7], calculate the power used over this interval for each wheel, and keep running sums or averages of wheel speed, power output, and distance. Wheel direction is sensed by using the computer's digital I/O port where the output from one of the opto-interrupters serves as the clock signal to a D-type flip flop and the output from

1 The WAFT prototype cannot presently accommodate specialized three- and four-wheeled track chairs because the front and rear wheel alignment of these chairs does not allow them to fit on the ramps.
the other (in quadrature) serves as the data signal [7]–[9]. The computer also provides timing for each exercise stage and maintains a summary data file for each exercise session.

In addition to the analog speedometers (Fig. 1(a)), the computer’s color monitor also provides feedback to the user. ASYST programs display both numeric and graphical information regarding speed, elapsed time, and projected wheelchair heading.

III. METHODS

A. Calibration of the MAGTURBOT™

For calibration, an unmodified MAGTURBOT™ shaft (with flywheel removed) was coupled through a universal joint to the power take-off port of a Harding Super Precision variable speed control lathe (Model HLV-H; Harding, Elmira, NY) via a rotary strain gauge torque sensor (Model 1253-450D; Sensotec, Inc., Columbus, OH) and displayed on a strip chart recorder and digital display. The lathe controller allowed for rotational speeds adjusted in 11 steps from 240 to 3500 r·min⁻¹. The seven MAGTURBOT™ settings were designated VVL (Very Very Light), VL, L, M (Moderate), H (Hard), VH, and VVH (Very Very Hard), corresponding to numerical settings 2 to 8 (setting 1 = FW; that is, “freewheel” or MAGTURBOT™ disengaged). These experiments were conducted to establish the braking characteristics of only the MAGTURBOT™ without the effects of adding the rollers and wheelchair.

B. Calibration of the Wheelchair Aerobic Fitness Trainer

For calibration experiments, a 73-kg automobile crash test dummy was seated in an 11.79 kg Quickie GP wheelchair (Motion Design, Fresno, CA) which was placed on the WAFT. Wheelchair tire pressure was measured before all experiments and was maintained at 4.14 N · m⁻². A 4-horsepower Servodisc™ DC torque motor (PMI Motion Technologies, Commack, NY) was coupled by a flexible universal joint to the wheelchair hub via the Lebow torque sensor, the output of which was conditioned and displayed on a strip chart recorder and digital display (Fig. 2). Shaft position was sensed by a capacitive rotary encoder (Trans-Tek Model 603) and displayed on a digital oscilloscope (Nicolet, Model 4094).

Speed was determined by capturing the encoded triangular wave form on the scope and measuring revolution time. Wheelchair wheel speed was also available from the WAFT’s speedometers which were also connected to a strip chart recorder (Figs. 1(a) and 2). A final indication of speed was given by counting via micro switch the number of wheel revolutions in a given time period. A Sorensen 45A/60V power supply, operated in a constant voltage mode, supplied the input to the torque motor. The torque motor speed was set by varying the output voltage of the power supply. Speeds in excess of 1 mph could be set accurately to within ±0.05 mph. Calibration experiments were conducted at speeds of 1, 2, 3, 4, and 5 mph since a range of speeds was needed to construct graded exercise protocols, and this range of speeds includes those observed in studies of manual wheelchair propulsion [10], [11].

The torque sensor was calibrated prior to and immediately after all calibration experiments. A 1.27-cm-diameter aluminum rod, grooved at 2.54-cm intervals along its length, was affixed to a quarter-inch socket drive and inserted into the Lebow torque sensor. A bench vice was used to anchor the Lebow torque sensor. With the aluminum rod in a horizontal position, a 4.5-kg weight was suspended from the rod at distances of 15.2 and 30.5 cm from the sensor to produce 6.8 and 13.6 N · m torques, respectively. Appropriate gain and offset adjustments were made and verified on a strip chart recorder to provide pen excursions ranging from zero (no torque) to 13.6 N · m full-scale (rated upper limit of the sensor).

During these calibration tests, the torque produced by any given combination of speed and MAGTURBOT™ setting was taken to be the average of the minimum and maximum chart recorder excursions. Differences between maximum and minimum displacement were on the order of 0.54 N · m. From a measure of torque and revolutions per min (r · min⁻¹), one can derive the power output (P) of a system [12]:

\[ P(W) = 2\pi \times T(N \cdot m) \times S(r \cdot min^{-1}) \times 0.0167(min^{-1}) \]

where \( P \) is power output calculated in W, \( T \) is torque measured by the strain gauge torque sensor, \( 2\pi \) is a constant, and \( S \) is the rotational speed in r · min⁻¹. From (1), combinations of observed speed and measured torque were used to compute a number of power outputs for each braking resistance setting.

C. Calculation of Power Output During WAFT Operation

As an initial method of determining power output curves for use during operation of the WAFT, a second-order polynomial least-squares regression equation was determined for each
resistance setting for each wheel using the power outputs computed over the given range of wheel speeds. By substituting the average wheel speed over a particular time interval into the appropriate equation (and assuming that speed does not vary appreciably over that interval), an “average” power output could be calculated. Total power output could then be determined by summing the power outputs of the left and right wheels.

Calculation of power output was further simplified by capitalizing on the linear relation found between resistance setting and power output. This linear relationship facilitated development of a single equation relating power output to both speed and resistance setting for each wheel as opposed to using a set of curves relating power output to speed for a single setting.

D. Arm Crank Ergometer versus WAFT: Graded Exercise Testing Protocols

Subjects’ upper body exercise performance on the new wheelchair ergometer was contrasted to their peak exercise tolerance on an arm crank ergometer. A Quinton CORIVAL-400 (Quinton Instrument Company, Seattle, WA) electrically braked ergometer was modified for arm cranking. The height of the ergometer was adjusted so that the midpoint of the chain-wheel was at shoulder level. During cranking, the subject’s arms, at point of maximal reach, were extended at right angles to the body and had a slight bend in the elbow. Special gloves were used to secure the hands of quadriplegic subjects to the hand cranks. A counter-clockwise cranking pattern was necessitated by the limited modifications that could be made to the ergometer. Average peak measures of oxygen uptake achieved with counterclockwise cranking did not appear markedly different from those reported in other studies for subjects of similar activity levels [13].

Arm crank ergometer exercise stages were 3 min in duration, with no rest intervals between stages. Subjects rested quietly for 15 min prior to beginning the exercise test. The initial workload was either 8 or 16 W, with increases of 8 or 16 W in subsequent stages [14], [15]. A preprogrammed Quinton Q3000B Stress Test Monitor automatically increased the workloads on the CORIVAL-400 ergometer every 3 min. Analog and digital displays of \( r \cdot \text{min}^{-1} \) and coaching by the investigators ensured that the subject held the prescribed cranking rate of 60 \( r \cdot \text{min}^{-1} \). Exercise stages for the WAFT were developed on the basis of power output data obtained from results of the calibration tests (see Results).

E. Physiological Measurements

Electrocardiogram and blood pressure were monitored for subject safety and as measures of exercise imposed cardiovascular stress. A standard electrocardiogram was recorded each minute and was continuously displayed for observation. Blood pressure was determined periodically by the auscultatory technique. For purposes of comparing arm crank and wheelchair ergometry, oxygen uptake (\( \text{VO}_2 \), \( \text{L} \cdot \text{min}^{-1} \)) was measured using the open circuit method (MMC Horizon™ System, SensorMedics, Anaheim, CA). Prior to and after each test, the analyzers were calibrated with reference gases and room air. A turbine volume transducer located within the MMC Horizon™ was used to determine minute ventilation, \( (V_E, \ L \cdot \text{min}^{-1}) \), tidal volume \( (V_T, \ L \cdot \text{breath}) \) and ventilation rate \( (V_e, \ \text{breaths} \cdot \text{min}^{-1}) \). Ratings of perceived exertion (RPE) were obtained using Borg’s 15-point graded category scale. Ratings were taken during the last 30 s of each workload and the last 30 s of the maximal workload.

F. Statistical Analysis

The strength of relationship between the dependent and independent variables was assessed by the Pearson correlation coefficient or coefficient of multiple correlation \( (r, R) \). Differences in peak exercise measures for subject groups completing arm crank ergometry (ACE) and wheelchair ergometry (WCE) were evaluated using a one-factor analysis of variance (ANOVA). Decisions regarding statistical significance were based upon an alpha level of 0.05. Mean values of power output and physiological measures are reported plus/minus one standard deviation.

IV. RESULTS

A. Calibration Experiments

1) MAGTURBO™: As shown in Fig. 3, a significant quadratic relationship was found between power output and rotational velocity for all settings of the magnetic turbo except the very, very low setting \( (r \geq 0.997 \) for all regressions; \( p < 0.01 \) for the quadratic term in all regressions except that for the VVL setting where \( p > 0.05 \). Of the seven turbo settings, four strata were seen in the relationship between speed and power: (a) VVL; (b) VL; (c) L, M, and H; and (d) VH and VVH. These strata ultimately had a marked influence on the formulation of graded exercise test protocols appropriate for a heterogeneous population of persons with lower limb disabilities.

2) WAFT Calibration: If the 2.9-cm-diameter shaft of the magnetic turbo was used to drive a 61-cm-diameter wheelchair wheel directly, a 21:1 ratio of diameters would result. As indicated in Fig. 3, wheel speeds in excess of 6 mph would then be required to operate the device in a range where power output is sensitive to variations in resistance setting. The belt
drive configuration actually used for the WAFT (Fig. 1(e)) increases the required speed still further. Since manual wheelchairs are commonly used at velocities of 2 to 3 mph [10], [11]; the MAGTURBO™ required some modification. By placing 4.0 mm brass shims between the housings that contain the two magnet arrays, the distance between the magnets was increased, and power output resolution at speeds within the range of normal manual wheelchair propulsion was greatly improved.

Power output curves for the WAFT with the MAGTURBO™ incorporated as described above are shown in Fig. 4(a). A significant quadratic relationship between power and rotational velocity for all settings of the magnetic turbo was found after it was integrated into the WAFT ($r > 0.997$ for all regressions and $p < 0.0001$ for all quadratic terms), and the stratification of turbo settings was still seen. No variation in power output was apparent between the two MAGTURBO™ units when tested alone. However, when the units were interfaced with the operating components of the WAFT, a small difference in power output between the right and left wheels was observed at all settings. As a result, separate power output equations for each wheel developed to provide precision in protocol development and measurement of subjects’ power output during wheelchair exercise testing. Minor differences in resistances between the sides did not have any apparent effect on the test performance of lower limb disabled subjects.

B. Calculation of Power Output During WAFT Operation

Though the relationship between power output and speed at each resistance setting is quadratic, the relationship between power output and setting at each speed is linear (Fig. 4(b)). To the extent that an identifiable relationship exists between wheel speed and the slopes and/or y-intercepts of the regression equations relating power output and resistance setting, both speed and setting may be incorporated into a single equation for calculating power output. Specifically:

$$W = \text{Setting} \times \text{Slope} \text{ (as a function of mph)} + \text{Intercept} \text{ (as a function of mph)} \quad (2)$$

Fig. 4(c) illustrates the significant quadratic relationship between speed and both slope and intercept of the linear power output/resistance setting relations ($p < 0.001$ and $r \geq 0.999$ for both regressions). Substituting these regression equations into (2), we obtain for the left wheel:

$$W_{L} = \text{Setting} \times (-0.02 + 0.519 \times \text{mph} + 0.301 \times \text{mph}^2) + (0.025 + 0.903 \times \text{mph} - 0.317 \times \text{mph}^2). \quad (3)$$

In the same way, a slightly different equation was derived for the right wheel:

$$W_{R} = \text{Setting} \times (-0.028 + 0.677 \times \text{mph} + 0.289 \times \text{mph}^2) + (0.052 + 1.072 \times \text{mph} - 0.111 \times \text{mph}^2) \quad (4)$$

It should be noted that this equation does not apply when speeds of 5 mph or greater are used in combination with MAGTURBO™ settings VH or VVH because under these conditions, slippage occurred between the wheel and the roller during calibration.

Total power output was obtained by summing the individual power outputs given by (3) and (4). Note that determination of total power output does not require that left and right wheels travel at the same speed or have the same resistance setting.

C. WAFT Exercise Protocols

Power output information from the calibration experiments was used to establish two graded wheelchair ergometer exercise test protocols. The low physical fitness (LPF) protocol was intended for the evaluation of patients with lower limb disabilities and symptoms of coronary artery disease, and/or for apparently healthy persons with low initial levels of physical fitness. The second protocol was designed for apparently healthy persons with average to above average initial levels of physical fitness and was called the average physically fit (APF) protocol. Both protocols were intended to produce an increase in metabolic equivalents (METS) consistent with the general principles for exercise testing proposed by the American College of Sports Medicine (ASCM) [16].
eight power output curves shown in Fig. 4(a), two protocols were established
for the WAFT. (a) Using six of the eight power output curves shown in Fig. 4(a), two protocols were established
by choosing combinations of speed and resistance setting to produce the desired stage-to-stage increments in power output. Here the two protocols,
and average physical fitness, are plotted on the power output versus speed curves for each MAGTURBO-TM resistance setting. (b) Total power output requirement (sum of right and left sides) at each stage of the two test protocols; power output increases linearly (or piecewise linearly) with stage ($r > 0.997$).

Both protocols are represented graphically in Fig. 5(a) and (b). In the LPF protocol, power output requirements began at 6 W in the first stage and increased by 5, 6, and 7 W [17]. Increases in exercise were accomplished by increases in magnetic braking resistance while speed was held constant at 2 mph. Stage 7 of the protocol called for a reduction of one resistance setting, with speed increased to 2.5 mph. In the APF protocol, increases in power output ranged from 5 to 20 W. Changes in power output were accomplished by a combination of increases in magnetic braking resistance and increases in speed over the range from 2 to 4 mph. A stage duration of 3 min was used in both protocols. It should be noted that few subjects in this study completed more than 4 or 5 exercise stages.

D. Physiological Measurements

Although it is not the primary purpose of this paper to present a report of physiological measurements recorded from exercise tests using these protocols, a few observations are necessary for further discussion of the operating characteristics of the new wheelchair ergometer. After giving their written consent, 72 subjects with lower limb disabilities completed one of the test protocols; oxygen uptake ($\text{VO}_2$, L min$^{-1}$) was measured for each. Nineteen of these subjects were classified as upper level injuries (ULI, C4-T3), 25 as middle level injuries (MLI, T4-T10), and 28 as lower level injuries (LLI, T11-L3+Fractures requiring wheelchair restriction for a period of 2 months). The majority of persons tested were sedentary and ranged in age from 17 to 69 years. Most of the subjects in the ULI group completed the LPF protocol. The peak power output for the ULI group averaged 26 ($\pm 16$) W. For these subjects, the average power output increase of 6 ($\pm 4$) W with ascending stages was in agreement with the 6W projected by the LPF protocol. The majority of subjects in the MLI and LLI groups were tested using the APF protocol. The average power output increase of 8 ($\pm 4$) W for these subjects was consistent with the 9W specified in the APF protocol. The peak power output for the MLI and LLI groups averaged 37 ($\pm 12$) and 39 ($\pm 16$) W, respectively.

Mean increases in METS with increasing stages were 0.5 ($\pm 0.3$) for tetraplegics and 0.7 ($\pm 0.4$) for paraplegics, in line with ACSM guidelines.

E. Physiological Measures Obtained by Arm Crank versus WAFT Exercise Testing

No significant difference in oxygen uptake was found between arm crank ergometry (ACE) and wheelchair ergometry (WCE) for subjects in the ULI ($n = 12$), MLI ($n = 21$), or LLI ($n = 21$) groups who completed both tests ($p > 0.05$) [13]. Mean peak VO$_2$ and power output are presented in Fig. 6 for all subjects in each experimental group at rest and at 50, 75, and 100% of peak VO$_2$ during WCE and ACE exercise. During wheelchair exercise, mean peak VO$_2$ occurred at power outputs of 26, 37, and 39 W, respectively, for ULI, MLI, and LLI groups. In contrast, power outputs achieved by all three groups with arm crank ergometry were more than double those obtained by wheelchair ergometry (57, 85, and 97 W, respectively, for ULI, MLI, and LLI groups), indicating the significant mechanical advantage of arm cranking over wheelchair propulsion.

Intergroup mean differences in power output were significant during ACE for ULI versus MLI ($p < 0.05$) and ULI versus LLI ($p < 0.01$), while contrasts between experimental groups for WCE were not. However, for both ACE and WCE, differences in mean peak VO$_2$ for all possible group comparisons were statistically significant ($p < 0.05$). Possible explanations for this finding include (a) greater inter-subject variability relative to economy of wheelchair propulsion than variability associated with severity of lower limb disability and (b) differences in the methods employed to produce a variable workload, i.e., 120 W at a given resistance and stage-dependent magnetic eddy current brake on the WCE versus the ACE electrical brake that is independent of crank r · min$^{-1}$.

V. DISCUSSION

Various factors associated with manual wheelchair propulsion introduce variability into the results of exercise tests performed on a wheelchair ergometer. These factors include (a) wheelchair design and quality of components parts, (b) environment in which the wheelchair is used, (c) extent of person's disability, (d) person's level of physical fitness, and
A. Instantaneous versus Average Speed and Power Output

The successful operation of the WAFT for assessment of cardiorespiratory fitness depends heavily on the willingness and ability of the subject to comply with the work demands specified by the test protocol. Since the resistance created by each MAGTURBOTM varies directly with the velocity of the wheelchair wheels, the ergometer does not have a mechanical or electronic system that automatically compensates when a subject fails to maintain the speed required for a given power output. Hence, if either the protocol or operating characteristics of the WAFT presented too great a psychological or physiological strain such that a subject would not or could not comply with the protocol, then little or no progressive stage effect would occur.

We may quantify the sensitivity of power output to variations in wheel speed for the left wheel by taking the derivative of (3):

\[ \frac{dW}{dmph} = (0.519 \times \text{Setting} + 0.903) \times (0.602 \times \text{Setting} - 0.634). \]

At 2 mph and a MAGTURBOTM setting of L (setting 4), the derivative is 6.5 W per mph. Thus, a ±0.1 mph deviation in speed results in a difference in power output of approximately ±0.7 W. In spite of this potential for power output variation, the WAFT appeared to provide a sufficiently stable load for exercise testing. In this investigation, subjects achieved protocol-specified power output increments during submaximal stages, but as expected, were unable to do so as they fatigued, approaching maximal exercise.

Fig. 7 presents a typical example of instantaneous and average speeds and power outputs produced during two exercise stages by a reasonably fit and active wheelchair user with a complete spinal cord injury (T7-8). The traces in Fig. 7 represent instantaneous values while horizontal lines indicate averages. Fig. 7(a) and (c) are for a 3 min exercise period, with the interval from 60 to 90 s replotted and expanded in Fig. 7(b) and (d). Under Stage 2 conditions (prescribed speed of 2.0 mph and a MAGTURBOTM setting of VVL or 2), the subject maintained an average speed within 3% of the target over the 3 min test period (Fig. 7(a), lower trace). The standard deviation in this subject’s speed under these conditions was within 10% of the target. At a higher resistance setting (VH or 7) and higher velocity (3 mph), the subject was approaching 90% of his peak power output and exhibited more variation in instantaneous speed (standard deviation of 0.5 mph or 17%). Still, the subject’s mean speed for the left wheel was almost exactly the specified 3.0 mph (Fig. 7(a), upper white line). For more fit subjects, a regular pushing rhythm was difficult to maintain at low speeds against a low resistance; for all subjects, greater pushing frequency was needed to maintain higher speeds against a higher resistance (Fig. 7(b)).

The instantaneous speeds shown in Fig. 7(a) and (b) can be converted to power outputs using (1). The resultant instantaneous and average power outputs are shown in Fig. 7(c) and (d). Results from an additional setting (L or 4) at a different speed (2.5 mph) are also included.

B. Validity of Wheelchair Ergometry for Cardiorespiratory Fitness Assessment

One objective in developing the WAFT was to provide a device for wheelchair exercise testing and aerobic conditioning of persons with lower limb disabilities who are physically fit or unfit, young or old. Glaser and Collins [18] have demonstrated that propelling a wheelchair over a tile surface at 3 km · h⁻¹ requires a power output of ~6 W and over a low-pile carpet requires ~16 W. Sawka et al. [19] presented evidence that the levels of exertion reported by Glaser and Collins equaled and exceeded the mean peak power output capacity of wheelchair-restricted persons 80 to 90 years of age. The work of these researchers was used to set nominal operating standards for the WAFT. For example, by starting with the WAFT in the freewheeling mode (MAGTURBOTM not engaged) and at a low pushing velocity (1.5 mph) and
in subsequent stages engaging the MAGTURBO\textsuperscript{TM}, it was possible to create a satisfactory multistage exercise test that reached a power output of 16 W and a velocity of 2 mph in four stages. Hence, multistage exercise testing with the WAFT can meet the needs of persons with very low cardiorespiratory fitness as well as assess the peak exercise tolerance of the highly conditioned person.

Evidence of the validity of the new wheelchair ergometer for evaluation of aerobic capacity is provided by Fig. 8(a) which relates the oxygen uptake on the WAFT to that obtained on the arm crank ergometer (y-intercept and slope not significantly different from 0.0 and 1.0, respectively, \( p < 0.05; r = 0.91; SE = 0.67 \)). Correlation coefficients and standard errors between peak measures of other selected parameters for continuous wheelchair and arm crank ergometry were \( V_{\text{E}} : r = 0.85; SE = 11.94 \), \( V_{\text{T}} : r = 0.90; SE = 0.24 \), heart rate: \( r = 0.95; SE = 9.49 \), and power output: \( r = 0.78; SE = 17.77 \) (see Fig. 8(b)). All of these correlation coefficients were significant \( (p < 0.001) \). In addition, 88 to 90% of the total variance between differentiated RPE for continuous WCE and ACE was accounted for in these analyses. This finding and the strength of the association observed between the criterion ACE exercise test and the experimental WCE was interpreted as empirical evidence that the new WCE and graded exercise test protocol provide a valid measure of both the physiological and psychophysiological factors that contribute to the peak functional capacity of persons with lower limb disabilities.

VI. SUMMARY

The WAFT, with modified MAGTURBO\textsuperscript{TM}, offers independent manipulation of speed and resistance for each wheelchair wheel. This flexibility makes the WAFT appropriate for graded exercise testing and conditioning of persons with lower limb disabilities who have widely varied exercise capacities. Changes in power output can be accomplished by (a) holding speed constant and varying the MAGTURBO\textsuperscript{TM} setting, (b) holding the MAGTURBO\textsuperscript{TM} setting constant and varying the speed, and (c) appropriately manipulating both speed and resistance. Further, these manipulations of speed and resistance can be carried out independently for either arm, thereby making allowances for persons whose disabilities have resulted in bilateral asymmetry. We have successfully conducted well over 100 maximal graded exercise tests on the WAFT, with subjects ranging from 17 to 69 years of age and possessing a wide range of lower limb disabilities and cardiorespiratory fitness [13].

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details of this device have been published in abstract form [5], [7]. [8]. This research was completed in partial fulfillment of the requirements for the Doctor of Philosophy degree in Physical Education at University of Wisconsin-Madison for W.E.L. [13].

REFERENCES


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