PERFORMANCE DIAGNOSTICS WITH INSTRUMENTED RACING WHEELCHAIRS: COMPARISON OF ATHLETES OF CLASS T52 AND T53

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The purpose of this study was to develop a cheap and flexible instrumentation system for on-track monitoring of velocity and associated parameters. Two wheelchairs of two top athletes (class T52 and 53) were equipped with a ferrite motor connected to the rear wheels and the back EMF (electromotive force) was converted to velocity after filtering and calibrating. Drag and rolling resistance were determined from the decreasing velocity when rolling freely and from a differential equation. In two 100 m races per athlete, the velocity, peak push acceleration, push frequency, inertial, drag and friction forces, peak push power and energy were calculated. Athlete 1 (T53) reached twice the maximum velocity and six times the average peak push power of Athlete 2, and produced 2.7 times as much energy over 100m. The system developed is useful for optimising race training.

KEY WORDS: wheelchair, instrumentation, mechanics, racing, disability, classification.

INTRODUCTION:

Wheelchair propulsion biomechanics has been investigated extensively (e.g. Vanlandewijck *et al.*, 2001), mainly using ergometers. Astonishingly, little data of on-track racing and elite wheelchair racers is available, which might be due to constraints in instrumentation. Nevertheless, both track (2 S-VHS camcorders; Chow and Chae, 2007) and wheelchair have been instrumented, in the latter either the push rim with strain gauges (Goosey-Tolfrey *et al.*, 2001) and 6-DOF transducers (Cooper et al., 1997; Dabonneville *et al.*, 2005; Three Rivers Holdings, 2008) or the rear wheels with velocometers (Moss *et al.*, 2003). At most, different paraplegics (class T53 and T54) have been compared so far (Chow and Chae, 2007), but no data exists on the comparison between para- and tetraplegics (T51 and T52). The loss of upper spinal and finger muscle function (T52) is reflected in the records of various sprint disciplines (100, 200 and 400m): class T52 is on average 23% (15-34%) slower than class T53 (calculated from: IPC, 2008). The aim of this study was to 1) develop a cheap and flexible system (hard- and software) for velocity monitoring of sprint training, and 2) to compare athletes of different classification (T52 and T53). Class T52 and T53 are defined as equivalent to complete cord injury at cord level C7-8 and T1-7 respectively (IPC, 2007).

METHOD:

Equipment and Instrumentation: We applied an instrumentation method comparable to the one developed by Moss *et al.* (2003). However, instead of using an optical encoder, we selected a ferrite motor (ACC337, Maxx Products International, Lake Zurich, IL 60047, USA), equipped with a toy wheel (diameter 5 cm), which can be connected to the frame of the wheelchair in various ways (Figure 1). The motor is driven by the rear wheels of the racing chair through the toy wheel, and the motor's back EMF is stored by a data logger (DI-710-ULS, DATAQ Instruments, Akron, OH, USA) at 1.2 kHz. We instrumented the wheelchairs of 2 test subjects: X-Limit (Subject 1) with AERO front wheel (Corima) and two DISC rear wheels (Corima, diameter 700 mm), and X-Limit S (Subject 2) with AERO front wheel (Corima) and two Ultima Track rear wheels (Panaracer, diameter 700 mm).

Test Subjects: *Athlete 1*: male, age 19, class T53, body mass: 50 kg, achievements: World Wheelchair and Amputee Games 2005: 3 silver medals, 3rd ASEAN Para Games 2005: 2 gold and 1 silver medal, International Paralympic Committee (IPC) Athletics World Championships 2006: silver medal (200m sprint) and Games record-holder (200m), 2006 ranked #1 in the world, Japan Paralympic Athletics Championship 2007: 2 gold medals in 100m & 200m. *Athlete 2*: female, age 31, class T52, body mass 42 kg, achievements: ASEAN Paragames 2008: 3 gold medals in 100m, 200m & 400m.

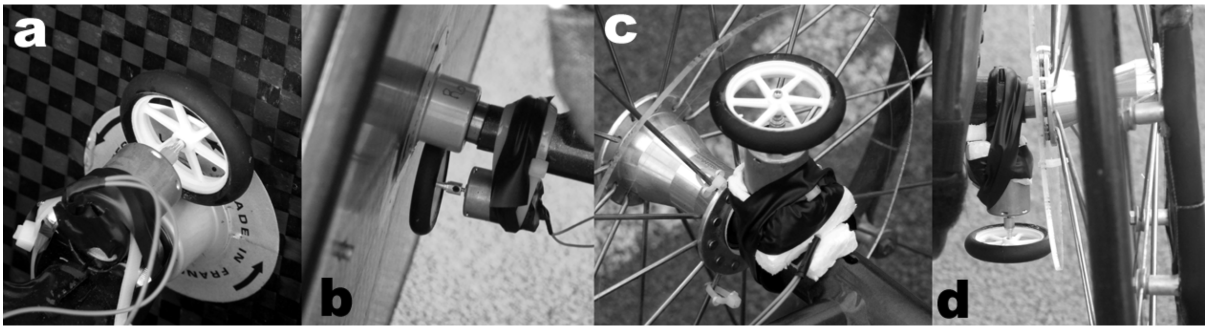

Figure 1: Instrumentation (a, b: wheelchair of Athlete 1, c, d: wheelchair of Athlete 2)

Calibration: The data stored in the data logger was uploaded with WINDAQ Waveform Browser (DATAQ Instruments, Akron, OH, USA). The first calibration of both wheelchairs was carried out on a conventional treadmill for exercising at velocities of up to 2 m/s, to test the linearity of the signal. The 2nd calibration test was performed on the track by covering a distance of exactly 100 m at various speeds (2-6 m/s). As the frequency of the noise is a function of the velocity, and as the maximal push frequency is close to the noise at lower speeds, we applied a 2nd order Savitzky-Golay filter thrice (window width: 501). The coefficient for converting the back EMF to velocity in m/s resulted from integrating the signal of the 2nd calibration tests, which must deliver 100m. The resulting mean coefficient produced an error of about 1% in the unfiltered signal, and about 0.1% in the filtered one.

Data Analysis: From the filtered velocity signal v of two 100m sprints per athlete, we calculated the acceleration a, the displacement s, the push frequency f, as well as the inertial force F_I and the kinetic energy E_{kin} of the racer–wheelchair system. The push frequency was calculated from the time between two subsequent peak accelerations. The dynamics of the acceleration and deceleration phases of the racer–wheelchair system are shown in Figure 2. The non-conservative energy of the system and the dissipative forces (drag and rolling resistance) were assessed in the following way: the wheelchair with the athlete in the typical aerodynamic racing position was accelerated to a certain speed and then allowed to roll freely until it came to a standstill. The experiments were performed at different speeds (2.5-6 m/s) in various directions. The data of free rolling, thereby decelerating, was fitted with a function, derived from the following non-linear differential equation:

$$m\,(dv\,/\,dt) + c_1\,v^2 + c_2 = 0 \qquad (1)$$

which is the force equilibrium of the inertial, drag and friction force, F_I, F_D, and F_R, of the deceleration phase (Figure 2), where m is the mass of the system, c_1 is the lumped drag coefficient (0.5 $\rho\, C_D\, A$, the product of air density ρ, coefficient of drag C_D and the frontal area A), and c_2 is F_R, the product of rolling friction coefficient and gravitational force (under the assumption that μ_r remains constant with v). The solution of Equation (1) is

$$v_1 = \left[\tan\left(\tan^{-1}\left(v_0\sqrt{\frac{c_1}{c_2}}\right) - \frac{t}{m}\sqrt{c_1 c_2}\right)\right]\sqrt{\frac{c_2}{c_1}} \qquad (2)$$

where v_1 is the instantaneous velocity decreasing with time when freely rolling, v_0 is the initial velocity at impending deceleration, and the coefficients c_1 and c_2, returned by the fit function, allow the separation of drag and rolling friction.

The horizontal force applied by the system to the ground, F_A, is the sum of F_I, the drag force F_D and the friction force F_R. F_A times v delivers the power input of the athlete through torque equilibrium and conservation of power, and its integration with time yields the overall energy input E_{in}. Subtracting the non-conservative energy E_R and E_D of rolling resistance and drag from E_{in} results in E_{kin}.

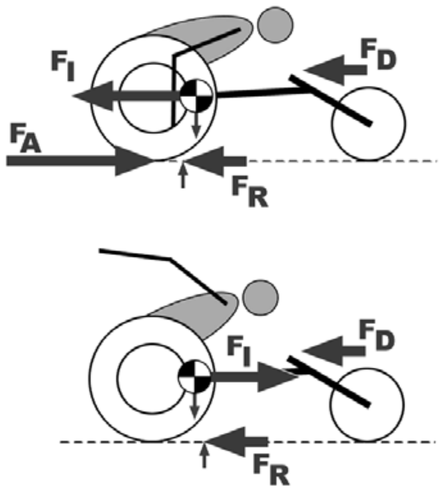

Figure 2: free body diagrams of acceleration (top) and deceleration phase (bottom)

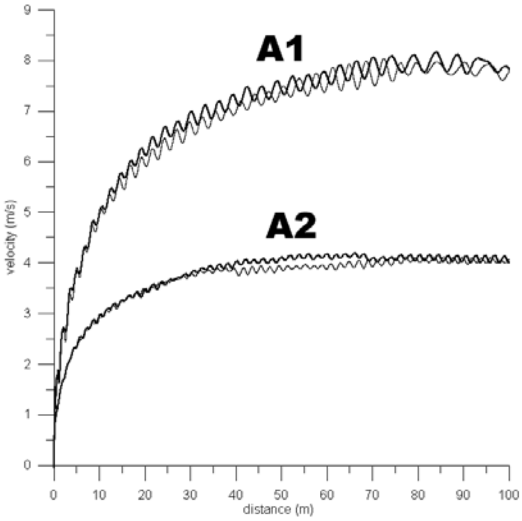

Figure 3: velocity profile, (A1 and A2 = Athlete 1 and 2)

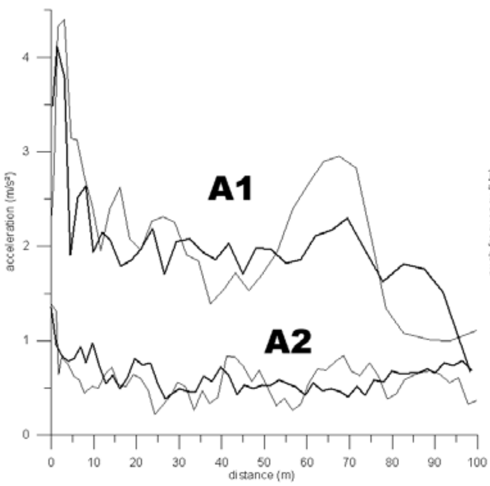

Figure 4: peak acceleration per push

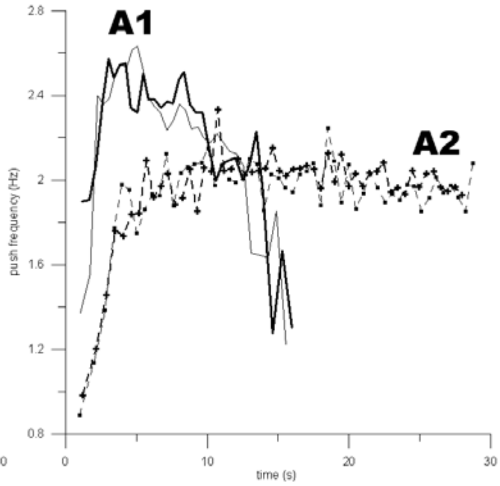

Figure 5: push frequency

RESULTS:

The velocity and acceleration graphs are shown in Figure 3 and 4. Athlete 1 reached a maximal velocity of 8 m/s, twice the one of athlete 2 (4m/s). Athlete 2 accelerated within the first 40 m before reaching a constant velocity. Athlete 1 accelerated throughout the first 75 m before reaching a constant velocity. Three distinct phases of acceleration can be distinguished: high acceleration (first 10m), less acceleration at higher push frequency (next 40m), and increased acceleration at lower push frequency (next 15m). At the constant velocity phase, both acceleration and push frequency decrease. The push frequency of Athlete 1 drops from 2.6 Hz to 1.3 Hz (Figure 5). In contrast to Athlete 1, the push frequency of Athlete 2 remains rather constant at 2 Hz after the acceleration phase. The peak push power of Athlete 1 and 2 is on average 900 W and 150 W per push respectively (Figure 6). Athlete 1 reaches a push power peak of 1.2 and 1.5 kW in the 3rd acceleration phase of the two 100 m sprints. The energy profile is shown in Figure 7. The total energy input over 100m is 3.5 kJ and 1.3 kJ in Athlete 1 and 2 respectively. The energy losses from drag and rolling resistance are equal in Athlete 1 at 100m, due to the higher final velocity.

DISCUSSION:

Chow and Chae (2007) reported a lower maximum speed in T53 athletes compared to T54. Their video method, however, does not reveal the acceleration of the stroke phases accurately, which are essential for calculating the inertial force. In contrast to Chow and

Chae's (2007) results, the stroke frequency was not constant in our T53 athlete, which rather dropped by 50% (Figure 5). Thus, wheelchair athletes do not necessarily, as Chow and Chae (2007) claimed, "prefer to maintain the same stroking rhythm when stroking with maximum effort", but rather develop their personal stroke pattern.

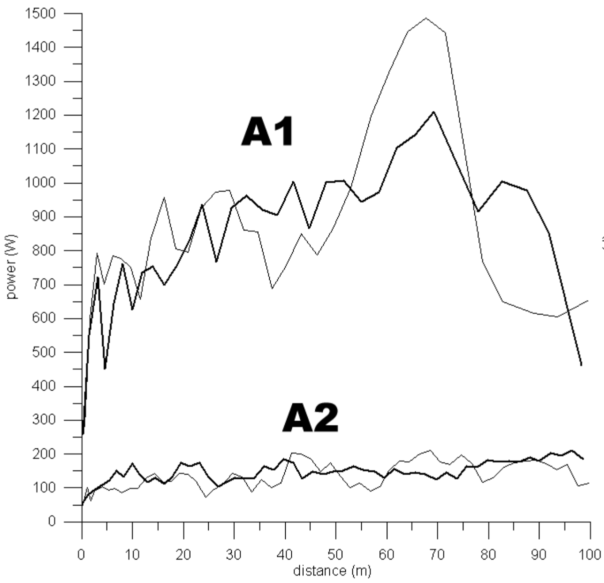

Figure 6: Peak power per push

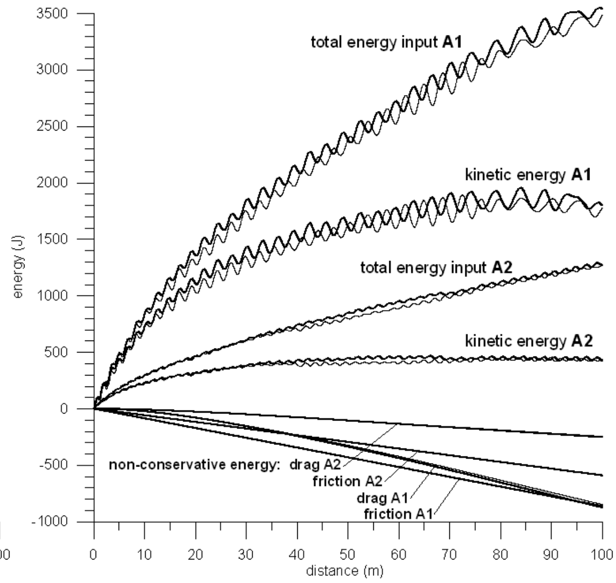

Figure 7: Energy profile over 100 m

CONCLUSION:

The wheelchair instrumentation method developed for this study provides a cheap and accurate system for velocity monitoring and wheelchair testing (drag and rolling resistance).

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