The effects of trunk kinematics and EMG activity of T54 wheelchair racing athletes on wheelchair propulsion speeds

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Background. The purpose of this study is to explore the influence of the trunk kinematic characteristics and trunk muscle EMG activity of wheelchair racing athletes on different racing wheelchair propulsion speeds. **Method.** The Vicon infrared high-speed 3D motion capture system was used to collect the kinematics data of the shoulders, elbows, wrists and trunk of twelve T54 athletes at four speeds (20 km/h, 25 km/h, 30 km/h and personal fastest speed) and the Trigno Wireless EMG system was used to collect the EMG (surface electromyography) data of the wheelchair athletes' rectus abdominis and erector spinus synchronous . The X-axis range of trunk movement of the wheelchair athletes at different propulsion speeds were compared using one-way ANOVA. The Pearson's correlation coefficient was used to analyze the correlations between the athletes' range of movement of the upper limb joints and trunk plotted on the X, Y, and Z axes. It was also used to analyze wheelchair propulsion speeds as well as the maximum angular velocity of the athletes' upper limbs and trunk on the X, Y, and Z axes and wheelchair propulsion speeds. The variables associated with wheelchair propulsion speed from the two sets of data were put into two multiple linear stepwise regressions. The Pearson's correlation coefficient test was performed on the EMG activity of the athletes' rectus abdominis and erector spinae muscles and wheelchair propulsion speeds. **Results.** There were significant differences in the athletes' raised angle, movement range and angular velocity of the trunk between wheelchair propulsion speeds of 20 km/h and the max speed ($P < 0.01$). The range of motion and angular speed of the athlete's trunk was significantly correlated with the propulsion speed of the wheelchair ($P=0.000$, $r=0.725$; $r=0.882$, $P=0.000$) and had a higher correlation coefficient than the correlation of the athlete's upper limb joints and the propulsion speed of the wheelchair. Multiple linear stepwise regression model results showed that the β values of the athlete's trunk motion range and angular velocity were the highest (β=0.484, β=0.422), and these values were also greater than the other independent variables in the two models. Most of the variables from the EMG of the athlete’s rectus abdominis were correlated with the propulsion speed of the wheelchair (p=0.000, r□0.5), and two of variables from the EMG of the athlete’s erector spinae were correlated with wheelchair propulsion speed (p=0.000, r=538; p=0.039, r=298).

Conclusion. The movement of the athlete's trunk has the largest effect on wheelchair propulsion speed. Athletes should use trunk movements to improve the running speed of the wheelchair and to avoid increased air drag caused by the excessive trunk raising. The findings of this study all suggest that T54 wheelchair racing athletes should add trunk strength training to their training plans and focus on improving the strength of the trunk’s flexion and extension muscles.

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Abstract

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Method. The Vicon infrared high-speed 3D motion capture system was used to collect the kinematics data of the shoulders, elbows, wrists and trunk of twelve T54 athletes at four speeds (20 km/h, 25 km/h, 30 km/h and personal fastest speed) and the Trigno Wireless EMG system was used to collect the EMG (surface electromyography) data of the wheelchair athletes' rectus abdominis and erector spinus synchronous. The X-axis range of trunk movement of the wheelchair athletes at different propulsion speeds were compared using one-way ANOVA. The Pearson's correlation coefficient was used to analyze the correlations between the athletes' range of movement of the upper limb joints and trunk plotted on the X, Y, and Z axes. It was also used to analyze wheelchair propulsion speeds as well as the maximum angular velocity of the athletes' upper limbs and trunk on the X, Y, and Z axes and wheelchair propulsion speeds. The variables associated with wheelchair propulsion speed from the two sets of data were put into two multiple linear stepwise regressions. The Pearson's correlation coefficient test was performed on the EMG activity of the athletes' rectus abdominis and erector spinae muscles and wheelchair propulsion speeds.

Results. There were significant differences in the athletes' raised angle, movement range and angular velocity of the trunk between wheelchair propulsion speeds of 20 km/h and the max speed ($P < 0.01$). The range of motion and angular speed of the athlete's trunk was significantly correlated with the propulsion speed of the wheelchair (P=0.000, r=0.725; r=0.882, P=0.000) and had a higher correlation coefficient than the correlation of the athlete's upper limb joints and the propulsion speed of the wheelchair. Multiple linear stepwise regression model results showed that the β values of the athlete's trunk motion range and angular velocity were the highest (β=0.484, β=0.422), and these values were also greater than the other independent variables in the two models. Most of the variables from the EMG of the athlete's rectus abdominis were correlated with the propulsion speed of the wheelchair (p=0.000, r＞0.5), and two of variables from the EMG of the athlete's erector spinae were correlated with wheelchair propulsion speed (p=0.000, r=538; p=0.039, r=298).

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Introduction

The position [1] and movement [2] of the athlete's trunk have a significant effect on wheelchair speed in wheelchair races, as trunk movement is one of the main ways to generate propulsive force[3]. Sanderson et al. published the first paper on the technical aspects of wheelchair racing [2], with the results showing that trunk movements significantly impact the force that is transferred from the trunk to the handrim and that the maximum force application point is on the handrim. The results of Goosey et al. were similar[4], finding that when athletes need to generate more force to propulsion, they not only need arm movement, but also need to lean their trunk forward to generate more momentum to the handrim. Leaning the trunk forward produces a large amount of vertical motion, contributing to acceleration[5]. Moss et al. showed[6] that the flexion of the trunk during wheelchair propulsion creates a favorable position for the upper limbs and hands to exert force on the handrim. This is because the leaning position of the trunk increases the angle of hand contact with the handrim in the driving phase[1, 5] and also increases the range of the whole propulsive phase[1, 7]. Therefore, the trunk movement of the wheelchair athletes plays an important role in improving propulsive force.

The posture and movements of a wheelchair athlete's trunk affects their propulsion technique[2, 6, 8, 9], but also affects their technical economy[10-13] and air drag[14-17], according to existing studies. Jones et al.[10] found that athletes with a higher technical economy have a better sense of rhythm and a lower trunk movement speed during propulsion. A study on the technical economy of wheelchair propulsion in wheelchair basketball players[12] found that reduced technical economy was related to an increase in trunk motion. In addition, trunk position affects the air drag the athletes receive in the process of propulsion[17], with air drag increasing after the trunk is raised[18]. This area of the trunk can be reduced by 0.17 m^2[17] when athletes adopt a more aerodynamic position, or by 18%[16] when using trunk flexion, thereby reducing air drag. If male wheelchair racing athletes adopted these techniques, they could improve their racing times by 116 seconds in a 5,000 m race[15]. By optimizing trunk posture and movement range, wheelchair athletes can improve their technical economy and reduce air drag, improving their competitive performance.

Trunk movements differ between different ages and genders of wheelchair racing athletes. Compared with young athletes[19] and female athletes[15], elite male athletes have a larger range of trunk movement. The trunk movements of wheelchair racing athletes with different class levels are also different[1, 20, 21]. Athletes with partial or no trunk mobility have a significant disadvantage in sports performance compared with athletes with full trunk mobility [22].

To date, only a few studies have used electromyography (EMG) to analyze muscle activation in wheelchair athletes during propulsion. Chow et al. used EMG to analyze the contraction characteristics of the upper limb muscles of wheelchair athletes using two different wheelchair racing techniques [7] and at different levels of resistance[23]. Umezu et al. investigated the differences in the EMG characteristics of the triceps of both elite and amateur wheelchair marathon runners[24]. These studies all focused on the EMG characteristics of the upper limbs of wheelchair athletes, but did not show EMG data about the trunk. At present, there is only one study on the EMG characteristics of the trunk of wheelchair athletes in a 400 m race[25], which showed that the EMG index of the rectus abdominis was only correlated with propulsion speed in the first 100 m, but not in the remaining 300 m of the race. This study also found that the back muscle group of the trunk was also not correlated with propulsion speed. A trunk EMG study observing daily wheelchair propulsion[26] found that the simultaneous contraction of the abdominal and back muscles during the starting stage improved the efficiency of the force acting on the handrim, and the activation degree of the trunk muscles increased significantly when the propulsion speed increased. These two studies are important to the body of research on the technical characteristics of wheelchair racing events, highlighting that the activation degree of the rectus abdominis and the erector spinae of wheelchair racing athletes in the process of propulsion may improve their performance.

According to the above studies, the movement and posture of the trunk have an important effect on a wheelchair athlete's competitive performance. Although there have been some studies on the kinematic characteristics of the trunks of wheelchair racing athletes, there are no studies on the kinematics of the trunk of wheelchair athletes at different speeds using EMG. The purpose of this study was to analyze the kinematics and EMG characteristics of the trunk of wheelchair athletes at different speeds. We hypothesized that the trunk movement of wheelchair athletes would have a major effect on the propulsion speed of the wheelchair and that the EMG activity of the rectus abdominis and erector spinae muscles would increase with propulsion speed.

Methods

Participants

Active athletes of wheelchair racing level T54 (with healthy upper limb and trunk functions), with no injuries and no drug or alcohol use before the trial were included in this study. A total of twelve athletes were enrolled, including ten male athletes and two female athletes, seven of whom were Chinese national team athletes who had participated in international events (Paralympic Games, Asian Paralympic Games), and five other athletes that had participated in national events (National Paralympic Games and National Championships). The longest training period of the athletes was 14 years and the shortest was 4 years. There were eight athletes in the kneeling posture, including four with polio, three with SCI (spinal cord injury) and one amputee. Four athletes were seated, including three amputees and one with polio. All of the athletes signed informed consent before the experiment, and this study was approved by the Ethics Committee of Shanghai University of Sport (IRB approval number: 102772021RT104). Table 1 shows the physical characteristics of the athletes included in the study.

Table 1. Physical characteristics of the wheelchair racing athletes included in the study

Instrumentation

The kinematic data were collected using the Vicon infrared High-speed Motion Capture system (T40) and 10 cameras (VICON Motion Systems, Oxford, UK) with a sampling frequency of 200 Hz. A total of 36 infrared, retro-reflective markers 14 mm in diameter were attached to the bone landmarks of the trunk and upper limb to define the trunk, shoulder, elbow and wrist.

The 16-wire EMG test system (Trigno Wireless EMG System, Delsys, Natick, MA, USA) was used for surface EMG data collection. The system includes a wireless EMG sensor (EMG signal width 20-450 Hz, signal sample rate 2,000 sample/s) and a base station. The sampling frequency used in this study was 2,000 Hz. EMG electrodes were placed in the rectus abdominis and erector spinus according to the positions suggested in the ABC of EMG[27].

Procedure

Warm-up: Each subject began by stretching their muscles and joints, including the head, neck, shoulder, elbow and trunk, then warming up in the wheelchair at slow speeds determined by each athlete (InvaCare, Top End, USA). The wheelchairs were mounted onto the wheelchair racing platform for the study (D&J, USA). All test instruments were checked during this 5-minute warm-up. A speedometer mounted onto the racing wheelchair was used to monitor wheelchair propulsion speed during the study.

Data collection

When the operator gave the instruction, the athlete began the propulsion, as shown in figure 1. Three relevant records for each speed were collected[2, 5, 7]. Athletes rested at 5-minute intervals between each speed to help prevent fatigue. A total of four speed tests, at 20 km/h, 25 km/h, 30 km/h and personal fastest speed were carried out, which are comparable to race and training speeds. The Trigno Wireless EMG telemetry system and the VICON system were used for the synchronous test, collecting the EMG and kinematic data of athletes, respectively. After data collection at all four speeds were completed, a maximum voluntary contraction test (MVC) was performed.

Figure 1. The scene of the technical test for wheelchair racing athletes

Data processing

The Nexus signal acquisition and processing software developed by VICON (VICON Nexus 2.6.1) was used to collect the kinematic signals with marker naming, noise removal, track deletion and other early signal processing carried out after data collection. The Visual3D analysis software (V3D, Version6, C-Motion Inc., USA) was used to calculate the kinematic data collected in this study.

EMG works 4.5 analysis software (DELSYS Inc., Natick, USA) was used to bandpass filter the collected EMG data (Butterworth, Passband width is 10-393 Hz). This software performed a baseline adjustment (removing the mean), full wave rectification, and a wave rectification to 1,000 Hz, and the final EMG data was then exported to a C3D file and imported into the Visual3D software (Version 6, V3D) to synchronize the EMG and kinematic data.

Based on previous studies, this study divided the technical movements of the athletes into three phases[7, 28, 29]—propulsion, release and recovery—and three specific time points—hand contacting the wheel, hand leaving the wheel and the highest point of the elbow. The propulsive phase starts when the hand contacts the handrim and continues until the hand releases contact. The release phase begins as soon as the athlete's hand leaves the handrim and continues until the elbow is raised to the highest point. The recovery phase is the period between the highest point of the elbow and the point at which the hand contacts the handrim again. In kinematic data, the X axis represents the flexion or extension of the joint in the sagittal plane, the Y axis represents the extension or adduction of the joint in the coronal plane, and the Z axis represents the external or internal rotation of the joint in the horizontal plane.

Analysis

IBM's SPSS Statistics 24 (IBM SPSS Inc., Chicago, IL, USA) was used for the statistical analysis in this study. The angle and range of movement of the trunk at different speeds were expressed by mean ± standard deviation. The Pearson's correlation coefficient test was used to analyze the correlation between propulsion speed and each joint's range of motion in the X, Y, Z axes at different propulsion speeds, as well as the correlation between the maximum angular velocity of each joint in the X, Y and Z axes and propulsion speed. The variables associated with propulsion speed ($P < 0.05$) in the two groups of data were used to establish a multiple stepwise regression model of each of the two groups with propulsion speed. The trunk motion angle and angular velocity between different speeds were compared using one-way ANOVA, and a Shapiro-Wilk normality test was conducted for each group of data before comparison. A Pearson's correlation coefficient test was conducted to determine the correlation between the EMG variables of the rectus abdominis and erector spinae with propulsion speed. Test results with a p value < 0.05 were considered statistically significant.

Results

Table 2. The angle and angular speed of the trunk (X-axis**) at different propulsion speeds**

The lean angle of the athlete's trunk (X-axis), the raised angle of the trunk (X-axis), the range of motion of the trunk (X-axis) and the angular velocity of the trunk (X-axis) at different speeds were compared using one-way ANOVA. The results showed (Table 2 & Figure 2) that there were significant differences at different speeds in the raised angle of the trunk (P=0.047) as well as the range of movement and angular velocity of the trunk (P=0.000，P=0.000).

Fisher's least significant difference (LSD) post hoc tests showed significant differences in rise angles between maximum speed and both 20 km/h and 25 km/h (P=0.009, P=0.027). There were extremely significant differences in the range of movement between maximum speed and the other three speeds (P=0.000, P=0.000, P=0.001), and between 20 km/h and 30 km/h (P= 0.001). There were also extremely significant differences in angular velocity between the maximum speed and the other three speeds (P=0.000, P=0.000, P=0.000), and between 20 km/h and 30 km/h (P=0.001).

Figure 2 The range of trunk movement of wheelchair athletes at different speeds

The correlation test between the range of motion and the propulsion speed showed that both the trunk (X-axis, P=0.000, r=0.725) and shoulder (X-axis, left: P=0.000, r=0.624; right: P=0.000, r=0.642) are significantly correlated with propulsion speed, with correlation coefficients greater than 0.6. The trunk on the X axis had the highest correlation coefficient of all variables tested (P=0.000, r=0.725). The left shoulder on the Y-axis (P=0.049, R=0.285) and the left and right shoulder joints on the Z-axis (left: P=0.024, r=0.326; right side: P=0.006, r=0.390) were also significantly correlated with propulsion speed. No other variables were correlated with propulsion speed.

Based on these results, the correlation between propulsion speed and the range of motion of: the trunk in the X axis, the left shoulder in the Y axis, and the left and right shoulder in the X and Z axis were all analyzed using multiple linear stepwise regression. These results are shown in Table 3.

Table 3 Multiple stepwise regression of propulsion speed to joint range of motion

The range of movement of the trunk on the X axis (X-Trunk), the right shoulder on the X axis (X-RSHO) and the left shoulder on the Y axis (Y-LSHO) were all entered into a multiple linear stepwise regression model. The overall regression model was statistically significant (p=0.000), explaining 60.5% of the propulsion speed ($^{\Delta}R^2$=0.605). The trunk had the highest $\boldsymbol{\beta}$ value on the X-axis (0.484), the right shoulder joint had the second highest $\boldsymbol{\beta}$ value on the X-axis (0.307) and the left shoulder joint had the lowest $\boldsymbol{\beta}$ value on the Y-axis (0.288).

Table 4 Correlation coefficients between propulsion speed and maximum angular joint velocity

Table 4 shows the results of the correlation analysis between the maximum angular velocity of the athlete's trunk, shoulder, elbow and wrist on the X, Y and Z axes and propulsion speed. All of the variables showed a significant correlation with propulsion speed except the maximum angular velocity of the left wrist on the Z-axis, which showed no correlation with propulsion speed. The trunk on the X-axis was the most significantly correlated with propulsion speed (r=0.882, P=0.000).

Table 5 Multiple stepwise regression of propulsion speed to maximum angular joint velocity

As shown in Table 5, the maximum angular velocities of the trunk (X-Trunk) and the left and right shoulder (X-RSHO, X-LSHO) joints on the X axis, the right shoulder joints on the Y axis (Y-RSHO), and the left elbow joints on the Z axis (Z-LELB) were included in the regression model. The model accounted for 87.2% of the wheelchair's propulsion speed. The maximum angular velocity of the trunk on the X axis (X-Trunk) had the highest beta value (β=0.422) of the variables included in the regression model.

Table 6 Correlation coefficients between propulsion speed and the integrated EMG of the trunk

Table 6 shows the correlation coefficients between the root mean square (RMS), and the integrated EMG of the trunk and propulsion speed. The integrated EMG of the trunk was taken at the following time points: the moment the hand contacts the wheel (P-RMS), the propulsive phase (P-INT), the moment the hand leaves the wheel (R-RMS), the release phase (R-INT), the moment the elbow is raised to the highest point (RE-RMS), the recovery phase (RE-INT), and the RMS maximum (RMS-MAX) during the entire movement. The integrated EMG of the rectus abdominis was correlated with propulsion speed in all time periods except in P-INT. The correlation coefficient was the highest when the elbow joint reached the highest point (RE_RMS, P=0.000, R=0.670). The erector spinae muscles were correlated with wheelchair propulsion speed only in the end of propulsive phase when the hand leaves the wheel (R-RMS) and at maximum RMS (RMS-MAX), but not in other time periods.

Discussion

This study analyzed the range of movement, maximum angular velocity and EMG activity of the trunk muscles of wheelchair athletes at different speeds. The results of this study are in line with the hypothesis that, compared with the shoulder, elbow and wrist joints, the movement of the trunk has the greatest influence on wheelchair propulsion speed, and the EMG activity of the trunk muscle is correlated with propulsion speed changes. In addition, the range of motion and angular velocity of the athlete's trunk gradually increase as the speed of the wheelchair increases.

A wheelchair athlete's posture and trunk movement impact race performance because the movement of the trunk in the sitting position not only generates optimal driving force, but is capable of optimizing the direction of that force[20]. The earliest study on the movements of wheelchair racing athletes found that the forward lean of the trunk can: increase power through the gravity of the body, increase the transfer of force from the trunk to the handrim, change the impact point of the force acting on the handrim, and reduce the reaction force of the handrim [2]. When the trunk is leaning forward and has a larger range of motion, it exerts its gravity onto the handrim through the rapid motion of the upper limbs, leading to faster acceleration in the propulsive phase[5]. During wheelchair propulsion, the flexion of the trunk also puts the shoulder joints, arms and hands in a more favorable position to exert power. Leaning the trunk more forward also increases the contact angle and contact range between the hands and the handle ring of the athletes[1, 6, 7, 20], creating a larger vertical work distance for the acceleration of the hands[5]. The movement and strength of the athlete's trunk are also very important for starting the wheelchair from a stationary state[3, 8]. A study on the kinematic characteristics of a wheelchair racing athlete starting from a static condition shows that[6] in order for a wheelchair in a stationary state to overcome inertia, the upper limbs need the help of the trunk to push the wheelchair into motion. Therefore, the trunk movement in the first propulsion to start is unique because the needs of a wheelchair in a stationary position are different than the needs of a running wheelchair. Moreover, the peak velocity of the wheelchair during the sprint start is related to the flexion and extension of the athlete's trunk[6].

The role and importance of these trunk movements may help explain differences in the performance of wheelchair racing athletes with different degrees of disability[8], different genders and different ages in wheelchair racing events. Some studies have shown that athletes with lower degrees of disability and higher levels of trunk functioning have a greater range of trunk motion[1, 9, 20, 21]. Compared with athletes with impaired or missing trunk functioning, athletes with full trunk functioning are able to provide stability for the power of the shoulder joints and upper limbs[1] and are able to use the trunk to generate more momentum on the handrim when pushing[29]. Studies have also shown that a weak trunk stability increases the wheelchair athlete's risk of shoulder joint injury[30, 31]. The gender, age and experience of the wheelchair athlete also impacts trunk movement, senior and elite male athletes participating in the Paralympic Games having larger trunk movements than female athletes[15] and young male athletes[19]. Elite male wheelchair athletes are also more likely to use larger trunk movements for acceleration[5]. The difference of trunk movement between different gender and different age perhaps due to different strength level and experience, the elite male athletes usually own better trunk strength and competition experience.

Based on these results, it is likely that athletes with higher levels of trunk strength and trunk motion range will have athletic advantages at the start of the race and during other periods of acceleration and sprint racing. This advantage is even more prominent in short distance events.

As shown in this study (Table 2 & Figure 2), although there was no statistical difference in the forward angle of athlete's trunk at different speeds ($P > 0.05$), the average values showed a gradually increasing trend, and there were significant differences in the raised angle of the athlete's trunk at maximum propulsion speed compared to both 20 km/h and 25 km/h ($P < 0.01$, $P < 0.05$). There was also a significant difference in the range of trunk motion at maximum propulsion speed compared to minimum propulsion speed (20 km\h; $p < 0.01$). These results indicate that the athletes were able to increase the working distance of the trunk by increasing the leaning angle and raised angle of trunk, thus affecting the propulsion speed of the wheelchair. The results of this study are consistent with the results of previous studies[5, 13, 32]. The multiple linear stepwise regression model established in this study shows that among all the joints of the body, the motion range and angular speed of the trunk have the greatest influence on wheelchair propulsion speed (the trunk has the highest standardized beta value in Table 3 and Table 5). Increasing the angular velocity of the athlete's trunk increases the momentum when the trunk has downward flexion, which increases the power exerted by the upper limbs on the handrim[5], and finally increases the propulsion speed of the wheelchair. This result suggests that for wheelchair athletes, increasing the angular velocity of the trunk may be one of the most important factors in maintaining maximum speed during a race, indicating that the movement and posture of the trunk are both very important to the propulsion speed of the wheelchair[1].

There are only a few studies using EMG characteristics to analyze the technical movements of wheelchair racing athletes. Chow et al. analyzed the EMG characteristics of the upper limbs of wheelchair athletes who used two different propulsion techniques[7] and different levels of resistance [23]. Umezu et al. studied the EMG characteristics of the triceps of elite wheelchair marathon athletes and amateur wheelchair marathon athletes[24]. None of the existing studies report the EMG characteristics of the trunk muscles of wheelchair athletes. There is only one previous research report on the correlation between the EMG activity of the rectus abdominis, iliocostal lumborum, pectoralis longissimus and trapezius of wheelchair athletes in a 400-meter all-out sprint[25], and the results show that only the rectus abdominis is correlated with wheelchair propulsion speed, and only in the first 100 meters. The middle trapezius had a negative correlation with wheelchair propulsion speed at 200 to 300 meters, while the iliocostal lumborum and the longissimus thoracis showed no correlation with propulsion speed.

The results of our study (table 6) showed that the rectus abdominis at P-RMS, R-RMS, R-INT, RE-RMS, RE-INT and RNS-max were all correlated with propulsion speed (p=0.000, r＞0.5). The erector spinae was only correlated with propulsion speed at R-RMS and RMS-MAX (R=0.538, P=0.000; R=0.298, P=0.039), which is similar to the findings of Kumnerddee et al. [25]. The correlation between the EMG activity of the rectus abdominis and propulsion speed at multiple time points indicates that maybe an increase in propulsion speed also increases the reaction force from the handrim when the athlete's trunk moves downward[2], increasing the demand for push power. Although this experiment was conducted in a laboratory environment, in real wheelchair races, more speed increases air drag on the bodies of the athletes, requiring more force to keep the wheelchair moving[28]. As the speed of the wheelchair increases, the athlete's trunk needs to activate more muscles to overcome the air drag as it flexes. It is worth noting that in a study on the correlation between the upper arm strength and sprint performance of wheelchair athletes[33], there was no correlation between upper body strength and sprint ability at 40 and 100 meters, though the author noted that the degree of disability of the athletes should be considered. In that study, the authors did not measure the trunk strength of the athletes, but based on the results and the inference of that study, it is possible that differences in trunk disability levels and strength levels of the athletes included in this experiment may be correlated with short-distance sprint ability.

The rectus abdominis was the only factor not correlated with wheelchair propulsion speed in P-INT (integrated electromyography during propulsive phase). This may be due to a high degree of recruitment and activation of the rectus abdominis muscle fibers from the beginning of the recovery phase to the propulsive phase, with the rectus abdominis having the highest correlation with propulsion speed at the beginning of the recovery phase, RE-RMS (the highest point of the elbow joint; p=0.000, r=0.670). The integrated EMG of the rectus abdominis during recovery (RE-INT) was also correlated with propulsion speed (p=0.000, r=0.568). It is speculated that the rectus abdominis accumulates the main power and momentum for the subsequent propulsion during the recovery phase, so the rectus abdominis is activated less during the propulsive process than during the recovery phase. In the propulsive phase, the rectus abdominus mainly relies on the momentum generated by the flexion of the trunk during the recovery phase and the strength of the arm to push the handrim. A study of able-bodied people pushing wheelchairs published a similar finding, with the study results showing that the trunk muscle groups had a high degree of activation in the early phase of propulsion[26].

Compared with the correlation between the integrated EMG of the rectus abdominis at various time points and propulsion speeds, the erector spinae was only significantly correlated with propulsion speed at R-RMS (p=0.000, r=0.538). In RMS-MAX, there was a weak correlation between the erector spinae and propulsion speed (p=0.039, r=0.298), but there was no correlation at any other time point. The high correlation between the integrated EMG of the erector spinae at R-RMS and propulsion speed indicates that at increased propulsion speeds, the activation of the erector spine muscles increases when the trunk is raised after the release phase, accelerating the movement frequency of the trunk. In addition, the above results show that with the propulsion speed increases, the athlete faces more resistance overcoming the downward flexion of the trunk than overcoming trunk raising. On one hand, athlete downward flexion the trunk will be more and more quick with the speed increases, on other hand, the wind resistance of the athlete's trunk will be increased[34]. So the resistance when trunk downward flexion will be increased, the athlete needs to activate more abdominal muscle fibers to meet the work requirements above. The main resistance during trunk raising comes from the weight of the athlete's upper body and the movement speed of the trunk. However, with increase in the movement frequency of the trunk, the posterior muscle groups (such as the erector spinae muscles) of the trunk elongate during downward flexion, storing the elastic potential energy of the muscles for the subsequent rise of the trunk[23], the resistance of raising is partially offset by the elastic potential energy stored by the elongated posterior muscle fibers. These reasons may explain the results of both this study and previous studies [25], compared to the high degree of correlation between the athlete's rectus abdominis and propulsion speed at various time points, the posterior muscle group of the trunk has a low level of the correlation with propulsion speed.

Excessive trunk raising can negatively impact the competitive performance of wheelchair athletes. First, an excessive range of trunk movement[11] as well as head and trunk movements that are too fast[10] reduce the economy of movement and increase oxygen consumption[4]. Second, different body postures affect air drag[28], which is important because it accounts for 35% of the total resistance[14]. Previous studies have shown that air drag gradually increases with an increase in propulsion speed[34]. When the propulsion speed reaches 6.97 m/s, air drag accounts for 46% of the total resistance. The effective surface area of the wheelchair athlete in the upright position with the trunk raised is higher than in the competitive position with the trunk flexed forward. Head position also affects the power output of wheelchair athletes by 2%[17]. According to the study results of Lewis et al., male athletes can save 116 seconds in a 5,000-meter race if they can optimize their trunk posture to be more aerodynamic[15]. Wheelchair athletes should also pay attention to excessive motion range and head position when using the trunk to propel the wheelchair, to maximize movement economy[11] and decrease air drag.

Conclusions

This is the first study to analyze the trunk kinematics and EMG characteristics of wheelchair athletes at different speeds in a single experimental condition. The results of this study are in line with the hypothesis that, compared with the shoulder, elbow and wrist, the movement of the trunk has the greatest influence on wheelchair propulsion speed, and the EMG activity of the trunk muscles is correlated with changes in propulsion speed. The trunk movements of T54 wheelchair racing athletes play an important role in both wheelchair acceleration and speed; increasing the movement range and speed of the trunk can significantly improve the propulsion speed of the racing wheelchair. Nevertheless, athletes need to keep the range of trunk movements reasonable, to avoid the increased air drag and decreased movement economy caused by the excessive raising of the trunk and head. Coaches and athletes should pay attention to the strength of the athlete's trunk. Adding core strength training, especially of the flexion and extension muscles of the trunk, into the strength training plan of T54 athletes may improve their racing performances.

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Figure 1

Test for wheelchair racing athletes' technique

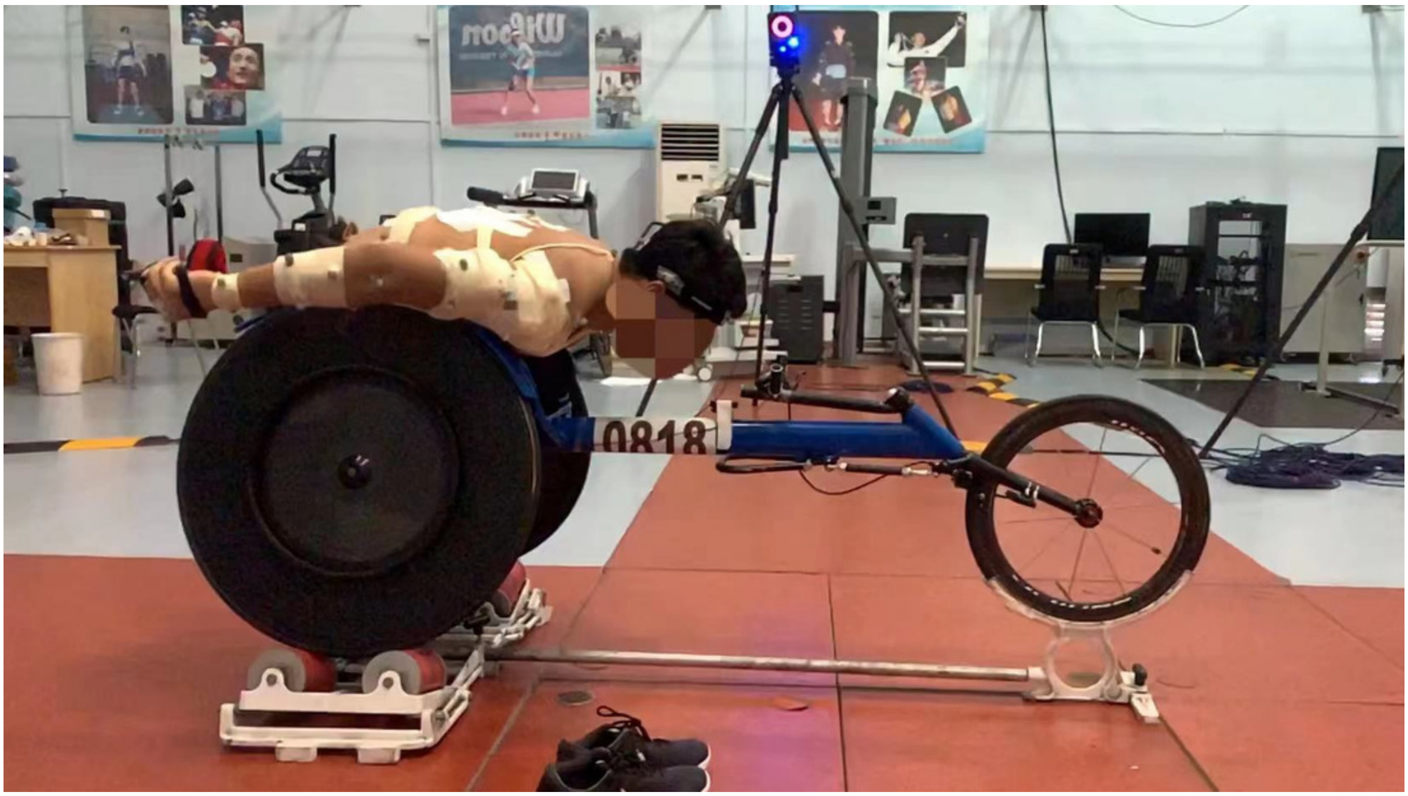

Figure 2

The range of movement of athlete's trunk at different speeds

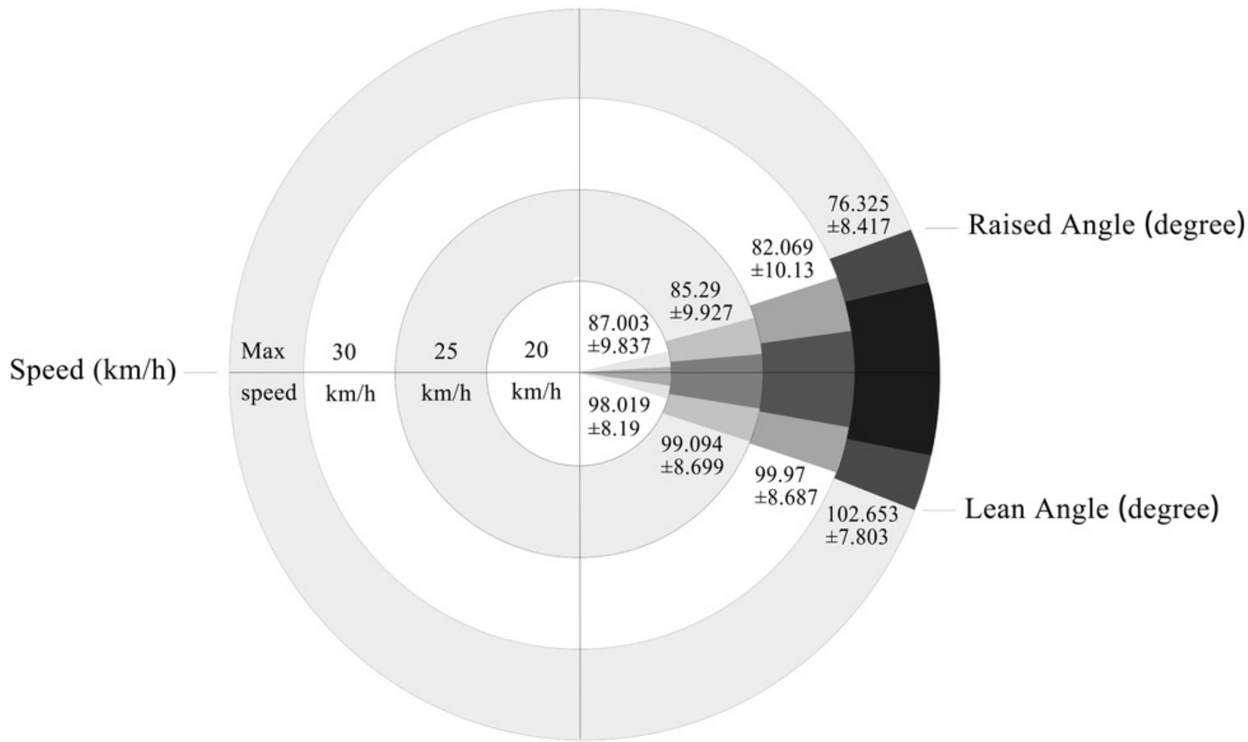

Table 1(on next page)

Physical characteristics of wheelchair racing athletes

Table 1 Physical characteristics of the wheelchair racing athletes included in the study

	n	**Age(*yr*)**	**Sitting Height（*cm*）**	**Weight(*kg*)**	**Years for training(*yr*)**
Man	10	21.7±4.22	89.7±6.13	79.9±9.68	6.7±2.87
Woman	2	28±5.66	83.5±0.71	72.3±6.9	9±7.07
Total	12	22.75±4.85	88.67±6.05	76.3±9.23	7.08±3.48

Table 2(on next page)

The angle and angular velocity of the trunk at different propulsion speeds

Table 2 The angle and angular speed of the trunk (X-axis) of wheelchair athletes at different propulsion speeds

Speed （*km/h*）	**Lean angle** （***degrees***）	**Raised angle** （***degrees***）	**Rang of movement** （***degrees***）	**Angular velocity** （***degrees/s***）
20	98.019 ± 8.190	87.003 ± 9.837**	11.02±4.47**	51.99±18.04**
25	99.094 ± 8.699	85.290 ± 9.927*	13.8±5.5	68.71±24.01**
30	99.970 ± 8.687	82.069 ± 10.130	17.9±6.48$^{\#}$	91.38±31.08**$^{\#\#}$
Max	102.653± 7.803	76.325 ± 8.417	26.33±4.05	180±29.37

*** (P < 0.05), ** (P < 0.01) denotes a significant differences with maximum speed; # (P < 0.05), ## (P < 0.01) denotes a significant differences with 20 km/h**

Table 3(on next page)

(on next page)

Multiple stepwise regression of propulsive speed to the range of motion of joint

Table 3 Multiple stepwise regression of wheelchair propulsion speed to joint range of motion

	β	T	p	VIF	ΔR^2	P
X-Trunk	0.484	3.490	0.001	2.287		
X-RSHO	0.307	2.205	0.033	2.298	0.605	P=0.000**
Y-LSHO	0.288	3.082	0.004	1.035		

Dependent Variable：pushing speed

Table 4 (on next page)

(on next page)

Correlation coefficient between propulsion speed and joint maximum angular velocity

Table 4 Correlation coefficients between wheelchair propulsion speed and maximum angular joint velocity

Axes		Trunk	Shouder		Elbow		Wrist	
			left	right	left	right	left	Right
X	r	0.882	0.844	0.814	0.531	0.828	0.494	0.551
	p	0.000**	0.000**	0.000**	0.000**	0.000**	0.000**	0.000**
Y	r	0.472	0.616	0.603	0.624	0.576	0.334	0.559
	p	0.001**	0.000**	0.000**	0.000**	0.000**	0.020*	0.000**
Z	r	0.493	0.467	0.703	0.539	0.442	0.101	0.588
	p	0.000**	0.001**	0.000**	0.000**	0.002**	0.496	0.000**

****（P＜0.01），*（P＜0.05）There was a significant correlation with pushing speeds**

Table 5(on next page)

Multiple stepwise regression of propulsion velocity to joint maximum angular velocity

Table 5 Multiple stepwise regression of propulsion speed to maximum angular joint velocity

	β	t	p	VIF	ΔR^2	P
X-Trunk	0.422	4.070	0.000	3.944		
X-LSHO	0.228	2.119	0.040	4.253		
Z-LELB	0.167	2.810	0.007	1.305	0.872	P=0.000***
Y-RSHO	0.164	2.422	0.020	1.689		
X-RSHO	0.163	1.708	0.095	3.334		

Dependent Variable：pushing speed

Table 6(on next page)

Correlation coefficients between propulsive speed and EMG of trunk

Table 6 Correlation coefficients between propulsion speed and the integrated EMG of the trunk

		P_RMS	P_INT	R_RMS	R_INT	RE_RMS	RE_Int	RMS_MAX
rectus abdominis	**r**	0.514	0.216	0.593	0.592	0.669	0.568	0.583
	p	0.000**	0.141	0.000**	0.000**	0.000**	0.000**	0.000**
erector spinae	**r**	0.197	-0.125	0.538**	-0.006	0.131	0.263	0.298
	p	0.181	0.396	0.000**	0.970	0.375	0.071	0.039*

**** （P＜0.01），*（P＜0.05）There was a significant correlation with pushing speeds**