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A Method for the Determination of Center of Gravity during Manual Wheelchair Propulsion in Different Axle Positions

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Abstract. The purpose of this study was to determine and verify a measuring and modeling technique for the center of gravity during manual wheelchair propulsion. Eight non-wheelchair users propelled themselves on linear flat floor and a slope (1/12) in four axle conditions. A three dimensional motion capture system and the force plates recorded static and dynamic trials of propulsion performance. Positions of the combined center of gravity of a wheelchair and the human body were calculated using a kinematic model. The positions of the center of gravity had a highly significant correlation with the positions of the center of pressure obtained from the force plate data in the anterior-posterior direction in static trials (r=0.99, p<.05), giving an average error (RMSE approximately 1.0). The minimal distance between the center of gravity and axle positions significantly decreased with axle position forwarded in propulsion on the floor (p<.05). Propulsion on the slope, however, demonstrated less significant differences of the distance between the center of gravity and axle positions. It implies that more dynamic activities lead to a variety of changes in the center of gravity. This method for determination of the center of gravity can be considered valid in static trials for wheelchair sitting and in dynamic trials on level ground during manual wheelchair propulsion.
Key words: Manual wheelchair propulsion, Center of gravity

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INTRODUCTION

Manual wheelchair propulsion is well-known as a locomotive form with low mechanical efficiency[1, 2]. High incidence of upper-extremity injury due to manual propulsion also has been reported by many researches[3, 4]. On the other hand, long-term use of manual wheelchairs is beneficial to maintain physical capacity because of engaging sufficient physical activity in daily life[5, 6]. Hence, improvement of propulsion efficiency is an essential need for manual wheelchair users and has come to be a common question in wheelchair biomechanics. A number of previous studies have addressed the effects of wheelchair design on propulsion efficiency. It is stated that a more forward axle position is correlated with more propulsion torque generation. Adjustments of rear wheels change the weight-bearing ratio of the rear wheels and the casters. The more forward the position of the wheel axis, the shorter the distance from the axis to the combined center of gravity of the wheelchair and user (COG). Consequently, a closer position of COG to the axis provides an increase in weight-bearing load on the wheels. It decreases a rolling resistance allowing users to generate more propulsion torque. However, a shorter distance between the axis and the COG results in less stability, which leads to backward tipping (wheelie) because of the more backward COG position on the wheelbase[7–9]. This tendency would become more critical in propulsion on a slope and /or over step in an obstacle environment.

This trade-off relationship has been well-documented in the past and is a key-factor in wheelchair design. However, very few or no studies have investigated the break-even point of torque generation and the COG position. That would limit information on wheelchair set-up process for users and professionals. Lemaire[10] developed a platform technique to determine the combined COG of the users and the wheelchair. Their calculation for the COG showed a very low measurement error. They presented the position of COG only in the anterior-posterior direction and in static trials, not in the vertical direction and in dynamic propulsion trials. Kobayashi[11] and Uchiyama[12] also measured the COG position in dynamic trials. They used a stationary instrument to propel the wheelchair, so subject performance tested was not a realistic style of wheelchair propulsion. A method for quantitative measurement of the COG during real wheelchair propulsion would contribute to get insight into the relationship between torque generation and changes in the COG position. Prior to examining the relationship between torque generation and the COG positions, a technique to measure positions of the COG will be needed. This study first aimed to construct a method to determine the COG position and to validate the method by testing on the COG position in different axle positions in static trials. The second aim was to investigate patterns of the COG trajectory during manual wheelchair propulsion on level ground and on a slope. The following hypotheses were tested. 1) The COG position calculated would be correlated with the position of center of pressure (COP) obtained from force plates in static trials. 2) The COG position calculated would correspond with changes in axle positions in static trials. 3) Trajectories of the COG position would show a typical pattern throughout propulsion by all subjects. 4) The COG position calculated would correspond with changes in axle positions in dynamic trials.

METHODS

Subjects

Eight participants were involved in this study, which was approved by the Institutional Review Board of the International University of Health and Welfare. Inclusion criteria were a non-wheelchair user and no problems and dysfunction in the upper-extremity and the trunk, and any physiological conditions of the circulation system. Table 1 presents the demographic characteristics of participants.

Instrumentations

A 12-camera three-dimensional motion capture system (Vicon612, ViconPeak) was used to collect the trajectory data of reflective markers on the subject bodies and the wheelchair at a sampling frequency of 120 Hz. Camera calibration was conducted for at least two cameras to see each marker throughout at least three full pushing cycles in a 4.0 m long ×2.0 m wide ×2.0 m high view volume. During propulsion trials on a slope, a ramp box was placed in the calibration volume. Six force plates (AMTI, USA) were used to obtain COP in static trials. The wheelchair used for data collection was Adapt (Inver care) with the tires size of 24 inches and the seat angle of 5 degree and the back-pipes angle of 5 degrees. The rear wheels were adjusted to at four positions (A, B, C and D) of the wheel axes (Fig. 1). The wheel axis was moved forward 2.5 cm each of the position in the direction of A-B-C-D. The seat height of each subject above the wheel axis was individually determined as the position in which a tip-finger reached the axis in sitting in the original (A) position.

Modeling

The biomechanical modeling for COG calculation involves the global coordinate and a wheelchair coordinate system. The global X and Y axes were horizontal, with the Y axes pointing in the progressional direction and the Z axis was vertical. The origin of the wheelchair local coordinate system was created at the midpoint between the right-and-left back-pipes at a height of the 38 cm from the floor level. The orientation of each of the three axes of the wheelchair local coordinate system was the same as that of the global coordinate system.

Center of pressure (COP) data from the force

Table 1. Subject characteristics

	Mean	SD
Years	26.5 yrs	± 4.5 (20–32)
Height	165.6 cm	± 9.1 (151–179)
Mass	60.9 kg	± 13.1 (42–75)

8 subjects (3 males, 5 females)

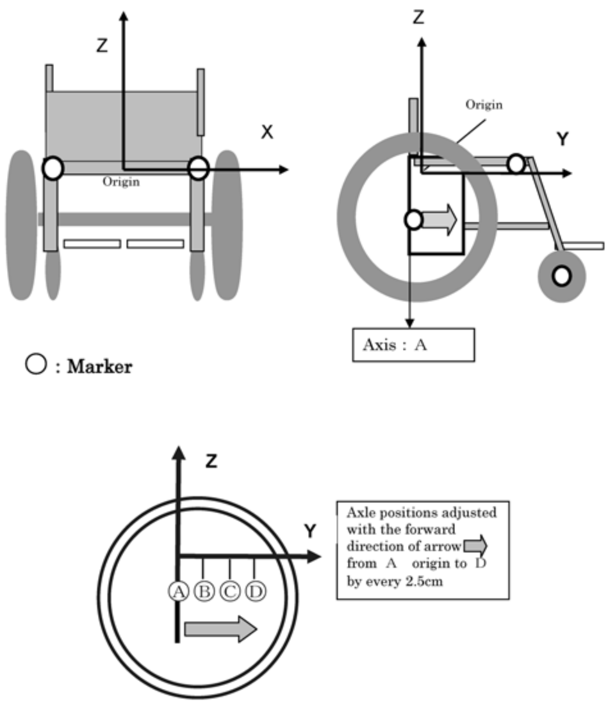

Fig. 1. Definition of wheelchair coordinate system and axle positions. Diagrams showing the wheelchair coordinate system defined by three markers on the seat frame and used to describe the wheelchair COG in four axle positions: A, B, C and D.

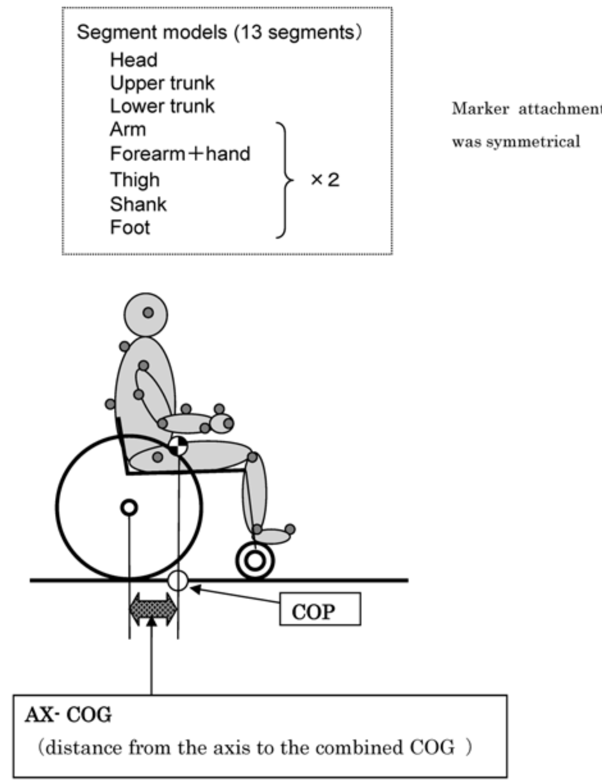

Fig. 2. Marker location on the body and segment model and Definition of AX-COG. The combined COG of wheelchair and body was calculated using segment models as presented above. AX-COG was defined as the distance between the axis and combined COG in the anterior-posterior direction. Values of COP were obtained from the force plates in static trials.

plates and position measurements from the motion capture system were collected to compute the positions of the wheelchair COG. The position of wheelchair COG in all the axle positions were calculated and filed as a parameter to be combined with positions of the body COG later. Body coordinate systems were modeled with thirteen body segments as rigid bodies in order to calculate a whole body COG (Fig. 2). Twenty-six 25 mm-diameter reflective markers were placed at the bony landmarks as illustrated in Fig. 2. The center of mass (COG) calculation for body segments was applied to the anthropometry parameters[13]. Each segment COG was combined to gain a body COG, which was combined again to determine the combined COG of the wheelchair and the human body.

Data collection: In order to obtain COP and weight-bearing load data from the force plates in each axle position, all subjects performed a fifteen-second static trial in two kinds of sitting posture: upright and relaxed postures with the hand gripping the top dead center of the handrim of the wheel. The subjects propelled wheelchair over the camera calibrated section of a smooth level carpet floor and wooden slope. Three trials were tested in each axle position both in floor and slope conditions. Subjects were instructed to push the wheelchair using their preferred technique, at a self-chosen comfortable speed on the floor, and at maximum-effort up the slope.

Calculation

Using the combined COG computed from the biomechanical models described above, the distance in the anterior-posterior direction between the axis and the COG was calculated and defined as AX-COG (Fig. 2). The shorter the AX-COG, the more unstable the COG position for users in the wheelchair. For propulsion on the floor, the COG position for AX-COG calculation was taken from the point that the COG moved most backward during trials. For propulsion on the slope (Fig. 7), an AX-COG was defined as the distance in the anterior-posterior direction between the ground contact point of the wheel and the most backward position of COG.

Statistics

Statistical analyses were performed with SPSS. Pearson correlation coefficient was applied to test the correlation between COP and COG in static trials. Analysis of variance (one-way ANOVA) was used to detect significant differences for the COG variables. Tukey tests were performed for multiple comparisons in axle positions. The level of statistical significance was decided as $p<.05$.

RESULTS

Validation of COG calculation in static trials

In the global coordinate system, Pearson correlation coefficient showed a high significant correlation between the Y values of the COP and COG positions ($p<.05$) (Fig. 3). The two kinds of static postures had the same level of correlation coefficient. The average errors between COP and COG were 3.5 mm (RMSE 0.83) and 4.7 mm (RMSE 1.09), upright and relaxed, respectively.

Axle positions vs. COG in static trials

A one-way ANOVA showed that axle position changes had significant effects on distance from the axis to the COG (AX-COG) both in the upright ($F(3,28)=25.5$, $p<.05$) and the relaxed ($F(3,28)=44.5$, $p<.05$) postures (Table 2). AX-COG decreased with axle positions adjusted forward in the direction of A to D. Furthermore, a significant difference in AX-COG was shown between the two kinds of static postures in the same axle position ($p<.05$). Weight-bearing load on the rear wheels increased with axle positions adjusted forward ($F(3,24)=14.1$, $p<.05$) (Table 2).

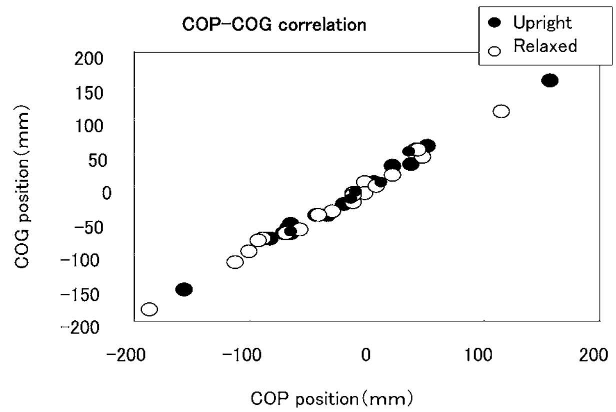

Fig. 3. Correlation between COP and COG. Values of combined COG were highly correlated with those of COP. Pearson correlation coefficients showed r=.99. RSME between COP and COG were 0.83 and 1.09, upright and relaxed posture, respectively.

The COG trajectory during propulsion

Trajectories of the COG position were like those during walking, repeating up and down shifting in the sagittal plane in the global coordinate system (Fig. 4a). Figure 4b presents an example of the COG trajectory of one pushing cycle during propulsion on the floor in the sagittal plane in the wheelchair coordinate system. All subjects demonstrated the same pattern of COG trajectory, in which the COG moved down and forward during the pushing phase (hand contact on the handrim) and moved up and backward during the recovery phase (hand off the handrim). Shapes of the COG trajectory slightly varied between the subjects. During propulsion on the slope, trajectories of the COG changes appeared to be the same pattern as that during floor propulsion, but the variability of trajectory shapes was greater than in floor propulsion both between and within subjects.

AX-COG during propulsion

The minimum AX-COG during floor propulsion showed significant differences between axle positions ($F(3,28)=20.5$, $p<.05$). A Tukey test detected significant differences between the axle A and B, C and D, and B and D ($p<.05$) (Table 2). For slope propulsion, significant differences of the minimum AX-COG (Fig. 5) were shown between axle positions ($F(3,28)=10.1$). A Tukey test showed the significant differences between the axes A and C, A and D, and B and D ($p<.05$) (Table 2). With

Table 2. Comparisons of axle positions in static trials and propulsion trials

	A	B	C	D
AX-COG static*	Mean SD	Mean SD	Mean SD	Mean SD
Upright (mm)	136.2 ± 16.2	113.2 ± 15.8	92.2 ± 15.8	71.0 ± 14.8
Relaxed (mm)	111.3 ± 10.4	95.1 ± 10.5	75.1 ± 10.4	56.1 ± 9.3
Ratio of weight bearing**				
Wheel (%)	69.4 ± 3.2	73.4 ± 3.1	76.4 ± 3.5	79.8 ± 3.9
Caster (%)	30.6 ± 3.2	26.6 ± 3.1	23.6 ± 3.5	20.2 ± 3.9
AX-COG propulsion***				
Flat floor (mm)	133.4 ± 17.6	108.6 ± 17.8	91.5 ± 16.8	69.8 ± 14.9
Slope (mm)	127.2 ± 26.3	116.1 ± 28.3	87.4 ± 17.6	69.5 ± 20.3

*Significant differences were found between all axle positions in both sitting postures. The COG in relaxed posture showed significantly shorter AX-COG than in the upright. Posture in all the axle positions and subjects (Tukey test, p<.05).
**Significant differences were shown between all axle positions (Tukey test, p<.05).
***AX-COG during propulsion decreased with changes of axle position: A to D. in floor propulsion, significant differences were found between A and B, A and C. A and D, B and D in floor propulsion. In slope propulsion, significant differences were found between A and C, A and D, B and D (Tukey test, p<.05).

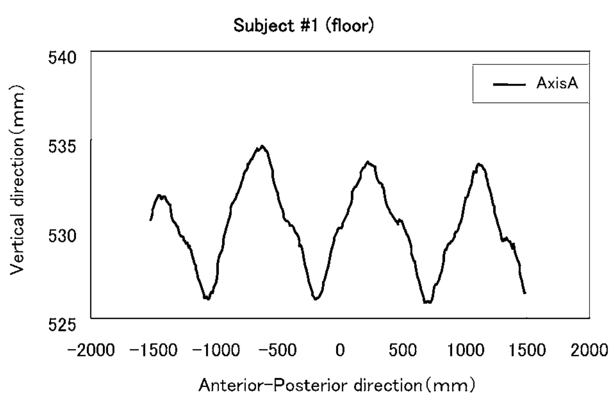

Fig. 4a. Trajectory of COG in the global coordinate system. An example of single subject during propulsion on floor.Trajectories of the COG were a wave-like cycle in the sagittal plane in the global coordinate system. The COG moves upward and downward repetitively.

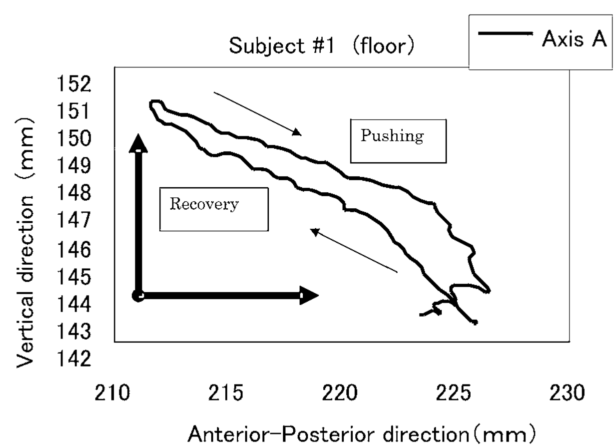

Fig. 4b. Trajectory of COG in the wheelchair coordinate system. An example of single subject during propulsion on floor. Trajectories of the COG were oval-like in shape in the sagittal plane in the wheelchair coordinate system. The COG moves forward and downward during the pushing phase, and moves back and upward during the recovery phase.

respect to shifting the distance of the COG in the anterior-posterior direction, it was ranged from 20 mm to 50 mm. Changes in axle positions were not related to the amount of COG shifting. In terms of the COG positions in the vertical direction, the highest position of COG showed no significant differences among axle positions.

DISCUSSION

A value of COP position must be identical with that of the COG position in the anterior-posterior direction in static trials. The results of validation for the COG calculation model demonstrated an extremely low error (approximately RMSE 1.0).

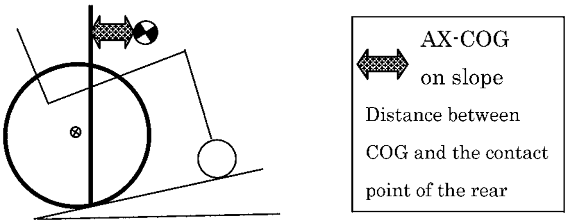

Fig. 5. Distance of AX-COG during slope propulsion. AX-COG for slope propulsion was defined as a distance between the ground contact point and combined COG in anterior-posterior direction.

Relaxed posture had a shorter AX-COG than upright posture. That means that the COG calculation determined a small change of posture. This model could, therefore, be considered valid for the determination of the COG in static trials. As a note, it is simply impractical to compare the COG with COP positions during dynamic trials. During propulsion, an assumption must be made that measurement errors are higher than 4.7 mm which was the average error shown in static trials.

In static trials, forward displacement of 2.5 cm in axle position resulted in a decrease of 2.2 cm in AX-COG on average. The COG calculation during static trials produced the expected result verifying a steady decline in the values of AX-COG as the axle position moved forward. Movements of the axle positions significantly affected the distribution of mass on the rear wheels and the front casters. The increase of 2% in weight-bearing load on the rear wheels resulted from moving the axle position forward by 2.5 cm. An inter-subject variance of 7–10% was observed in the weight-bearing ratio of the wheels and casters. The mass of the wheelchair did not change for all subjects, but the weight-bearing ratio of the wheels was not related to the mass of the subjects. Even in the same axle position, inter-subject variances occured in weight-bearing ratio of the wheels.

In the wheelchair local coordinate system, the trajectory of COG drew an oval-like figure with repeated backward and forward shifting of 2–5 cm in the anterior-posterior direction. The most backward position of the COG appeared to be at the initial time of the hand on the handrim, and the most forward position was at the time of the hand off the handrim. This pattern of the COG change would be associated with a range of motion of the trunk and the shoulder joint in flexion-extension.

While the AX-COG during static trials had theoretical and significant differences between all the axle positions, the AX-COG during floor propulsion did not. There were no significant differences between axes B and C, C and D in the multiple comparison. Propulsion activities altered the postures of the subjects, thus the COG movements became more dynamic. In terms of slope propulsion, it was the case that the AX-COG did not always decrease when the axle position moved forward in the direction from A to D. The COG movements may be affected by more dynamic posture sway and control required in slope propulsion than in floor propulsion. With respect to the COG change in the vertical direction, it was unaffected by change in axle positions. That may have been due to the other properties such as body height and/or propulsion styles.

Our experiment contains many limitations. The subjects participated were not actual wheelchair users who will have fundamentally different characteristics of propulsion movement. As a second limitation, there are no empirical techniques to compare the COG position with COP position in the vertical direction. It cannot be stated that this COG calculation model has valid applications for research which tests the determination of the COG. In addition, there were no approaches to validate the COG position during dynamic trials. The other limitation was that adjustments of wheelchair set-up were made only to axle position. Seat angle may also have an important effect on COG changes and be an effective variable for determination of COG modeling method.

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