Effects of Wheelchair Seat-height Settings on Alternating Lower Limb Propulsion With Both Legs

Tomoyuki Murata PhD\(^{ab}\), Toyoko Asami MD\(^{bc}\), Kiyomi Matsuo\(^{c}\), Atsuko Kubo PhD\(^{d}\) & Etsumi Okigawa\(^{a}\)

\(^{a}\) Kanagawa Rehabilitation Center, Kanagawa, Japan
\(^{b}\) Graduate School of Medicine, Saga University, Saga, Japan
\(^{c}\) Rehabilitation Center, Saga University Hospital, Saga, Japan
\(^{d}\) Nishikyusyu University, Saga, Japan

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Effects of Wheelchair Seat-height Settings on Alternating Lower Limb Propulsion With Both Legs

TOMOYUKI MURATA, PhD1,2, TOYOKO ASAMI, MD2,3, KIYOMI MATSUO3, ATSUKO KUBO, PhD4, and ETSUMI OKIGAWA1∗

1Kanagawa Rehabilitation Center, Kanagawa, Japan
2Graduate School of Medicine, Saga University, Saga, Japan
3Rehabilitation Center, Saga University Hospital, Saga, Japan
4Nishikyusyu University, Saga, Japan

This study investigated the effects of seat-height settings of wheelchairs with alternating propulsion with both legs. Seven healthy individuals with no orthopedic disease participated. Flexion angles at initial contact (FA-IC) of each joint, range of motion during propulsion period (ROM-PP), and ground reaction force (GRF) were measured using a three dimensional motion capture system and force plates, and compared with different seat-height settings. Statistically significant relationships were found between seat-height and speed, stride length, knee FA-IC, ankle FA-IC, hip ROM-PP, vertical ground reaction force (VGRF), and anterior posterior ground reaction force (APGRF). Speed, hip ROM-PP, VGRF and APGRF increased as the seat-height was lowered. This effect diminished when the seat-height was set below −40 mm. VGRF increased as the seat-height was lowered. The results suggest that the seat-height effect can be attributed to hip ROM-PP; therefore, optimal foot propulsion cannot be achieved when the seat height is set either too high or too low. Efficient foot propulsion of the wheelchair can be achieved by setting the seat height to lower leg length according to a combination of physical characteristics, such as the user’s physical functions, leg muscles, and range of motion.

Keywords: manual wheelchair, motion analysis, wheelchair propulsion, wheelchair seat-height setting

Introduction

To assist with walking, people with gait disturbances often use assistive devices, such as handrails, canes, walkers, and wheelchairs. The choice of device usually depends on the degree of gait disturbance. Wheelchairs can be categorized into three types: manually propelled wheelchairs, electric wheelchairs, and electrically powered scooters (Japanese Industrial Standards Committee, 2006a,b; 2009). Methods of operation of manually propelled wheelchairs include: wheelchair propulsion with hands, simultaneous propulsion with both legs, alternating propulsion with both legs, propulsion with hands and legs, propulsion with one leg, and propulsion with one hand and one leg. A precedent study reported that gait disorders and falls are closely related in older members of the population (Yves, Stephanie, & Reto, 2010). Wheelchair propulsion using the legs provides a safe mode of moving with a low risk of knee buckling or falling. It is generally chosen for those with hemiplegia, rheumatism, or incomplete spinal cord injury, and for the elderly. When seated in a wheelchair, the weight normally applied to the legs can be relieved, but there is still the ability to move using the residual function of the legs (Stein, Chong, James, & Bell, 2001); along with maintenance and improvement of mobility and muscle strength, and prevention of bone density loss (Genda & Tanaka, 2008; Matsuse et al., 2006). Wheelchair use can therefore be considered a tool which helps to provide an active life (Engström, 2003).

However, propelling a wheelchair becomes difficult owing to methods of adaptation. For example, if body measurements, physical function size or placement of the drive wheel are incompatible, then propulsion efficiency decreases (Peter, Jean, & Denise, 1994; Sean, Rick, Mochael, & Rory, 1998). Such mismatches in hand wheelchair propulsion units are reported to have caused secondary disabilities such as scoliosis and joint pain in the arms (Gallmann et al., 1988; Pentland & Twomey, 1994a,b; Sie, Waters, Adkins, & Gellman, 1992). It is therefore important that the correct wheelchair is chosen and adjusted to an individual user’s characteristics to prevent secondary disabilities with long-term use. When choosing and adjusting a wheelchair, the environment in which the user will predominately be using the wheelchair in, individual body measurements, physical functions, and the three functions of wheelchairs (moving, seating position, and transferring) need to be considered (Engström, 2003; Cooper, 1998).

To provide insight into causes and prevention of upper limb injuries in manual wheelchair users, pushrim forces and hub moments occurring during wheelchair propulsion have been analyzed (Boninger, Cooper, Rick, & Shimad, 1997). There have been studies on the positional relationships between the hands and drive wheel (Louise, Mario, Michael, & O’Riai, 1992); and between the hands and hand rim for hand-propelled wheelchairs.
(Hughes, Weimar, Sheth, & Brubaker, 1992); and on mechanical factors, such as the position of the center of mass and the wheelbase of the wheelchair (James, 2000; Kersti, Hans, Eva, & Bjorn, 2004), and the camber (Dirkjan, Luc, & Rients, 1989; Trudel, Dirkjan, Luc, & Rients, 1997) and caster angles of the drive wheel (Engström, 2003). It has also been reported that a change in wheelchair seat position caused more variation in integrated electromyography of for the triceps brachii, pectoralis major, and deltoid posterior (Louise et al., 1992). Providing a wheelchair user with an adjustable axle position and correctly fitting the wheelchair to the user’s characteristics can improve propulsion biomechanics and likely reduce the risk of injury (Boninger et al., 2000).

For wheelchair with one-leg propulsion, it has been reported that the driving force increases when seat height is set at low (Arima, Morioka, Hashizaki, & Nakamura, 1990). For wheelchairs with one-hand-one-leg propulsion, it has been reported that seat height and depth affect driving speed and ease of operation of the wheelchair (Yamada, Eguchi, Okayama, Okyama, & Uematsu, 1992). There are also reports on the relationship between lower leg length and seat height that is set in three stages for propulsion with both legs. Improper seat height setting is reported to lead to poor moving speed (Saito, Matsumoto, & Yoshinaga, 2006). However, there have been no reports on how manipulation of seat-height influences wheelchair propulsion with both legs.

In the current study, the effects of seat-height settings using healthy individuals using a wheelchair, propelled by alternating legs were analyzed. The objective was to investigate the effects of seat-height settings of wheelchairs with alternating propulsion with both legs, with the aims of providing useful information on seat-height settings. Data were collected from tests using a three dimensional motion capture system and force plates.

Materials and Methods
To study the effects of wheelchair seat-height settings on alternating lower limb propulsion with both legs, measurements were taken using an adjustable wheelchair (Basic; Etac, Sweden) that could be matched to a subject’s body measurements. From information provided in “How to adjust a wheelchair” (RESJA Wheelchair SIG, 2013), effective seat depths and widths were adjusted to match body dimensions, buttock/thigh depth, and pelvic width. Wheelchair seat heights were set at the following seven levels for each subject: (1) +60 mm, (2) +40 mm, (3) +20 mm, (4) ±0 mm, (5) −20 mm, (6) −40 mm, and (7) −60 mm. A subject’s heels were raised off the ground when the seat-height was set at +20 mm, +40 mm, and +60 mm in relation to leg length. A seat-cushion (20 mm thick) on top of a 5-mm thick board was used to minimize any deflection effect from the seat. The seat angle was set at 0° during measurements.

Subjects
Participants were seven healthy individuals with no orthopedic disease (4 males and 3 females; aged 31.1 ± 4.7 years; height: 167.4 ± 7.0 cm; mass: 60.6 ± 10.8 kg; lower leg length: 42.6 ± 5.5 cm). Measurements were taken after the aim and content of the study, and the manner in which personal information was to be treated, had been described, and participants had given their informed consent.

Instruments
A three dimensional motion capture system (Vicon MX Workstation 5.2.4; VICON, UK) synchronized with digital video cameras, and four force plates (BP400600-4000; AMTI, USA) were used for measurements. A measurement space of 4 m long, 3 m wide, and 3 m high was calibrated with 10 cameras, and each setting was tested by performing propulsion cycles on the force plates from an arranged starting position at 1 m from the force plates. The propulsion cycle was between initial contact and the beginning of the initial swing. For each person, three samples per seat-height setting were taken. The value of one sample represents the average of four cycles in one trial. Subjects were told to propel wheelchair at a comfortable speed. Ground reaction force (GRF) measurements were taken to avert the effect of weight of the wheelchair. Positional data obtained from measurements were imported to a computer at a sampling frequency of 60 Hz

Data Collection and Analysis
The propulsion period was defined as the time that each foot was placed on the floor. The coasting period was defined as the time that each foot was lifted from the floor (Tanaka, Ito, & Iizima, 1982). Speed, stride length, and cadence, which are indicators of gait, were calculated using a behavior analysis program (Body
Effects of Wheelchair Seat-height Settings on Foot Propulsion

Infrared reflective markers were attached to a participant’s body at 35 locations, which included the greater trochanter, with reference to the plug-in-gait model.

Fig. 2. Infrared reflective markers were attached to a participant’s body at 35 locations, which included the greater trochanter, with reference to the plug-in-gait model.

Fig. 3. The average values for speed, stride length and cadence for each seat-height setting are present. *One-way ANOVA, effect of seat height setting (p < 0.01).

Builder, 3.6; VICON, UK) from ground force reaction data and markers positioned on the heel and second metatarsal head. The timing of initial contact was calculated by the value of the force plate gauge. Differences between joint angles of the trunk, hip joint, knee joint and ankle joint were calculated by flexion angles at initial contact (FA-IC) and the range of motion during the propulsion period (ROM-PP). These values were calculated from data from one leg using each individual subject’s posture while standing still as a reference, and the average from four propulsion cycles as the value for each setting. GRF was sub-grouped into horizontal ground reaction force (HGRF), anterior posterior ground reaction force (APGRF) and vertical ground reaction force (VGRF), then comparisons and examinations of changes based on seat height settings were conducted. APGRF was the shearing force in the anterior posterior direction. VGRF was the frictional force generated between the foot and the floor. Resultant force of APGRF and VGRF affected the propulsion force of the sagittal plane (Jacqelin, 1992).

Statistics

A one-way analysis of variance (ANOVA) was used for seat-height settings for speed, stride length, cadence, each joint FA-IC, ROM-PP, and relationship with GRF. Correlations between items were analyzed using Spearman’s rank order correlation coefficient. A p value of less than 0.05 was considered statistically significant.

Results

The average values for speed, stride length and cadence for each seat-height setting are presented in Figure 3. ANOVA results revealed significant differences between seat height and stride length (p < 0.01). Statistically significant relationships were found between seat height and speed (p < 0.01, r = 0.73), seat height and stride length (p < 0.01, r = 0.89), and speed and stride length (p < 0.01, r = 0.86). No statistically significant relationships found between seat height and cadence, speed and cadence, or stride length and cadence (Table 1).

ANOVA results revealed significant differences between seat height and trunk FA-IC (p < 0.01), knee FA-IC (p < 0.01), and hip ROM-PP (p < 0.01; Table 1). The average values for trunk FA-IC, hip FA-IC, knee FA-IC, ankle FA-IC, trunk ROM-PP, hip ROM-PP, knee ROM-PP and ankle ROM-PP for each seat-height setting were compared. Statistically significant relationships were found between seat height and trunk FA-IC (p < 0.01, r = 0.35), seat height and knee FA-IC (p < 0.01, r = −0.56), seat height and ankle FA-IC (p < 0.01, r = −0.62), seat height and trunk ROM-PP (p < 0.01, r = 0.29), and seat height and hip ROM-PP (p < 0.01, r = 0.75; Table 1).

The average values for VGRF, APGRF and HGRF for each seat-height setting are presented in Figure 4. ANOVA results revealed significant differences between seat height and VGRF (p < 0.01) and APGRF (p < 0.01).

Discussion

Normal gait stance begins with an initial contact of the heel to the ground and terminates with contact with the toe (Kirsten, 2006). Optimal foot propulsion occurs through a normal gait pattern beginning with heel strike, mid-stance, and ending in the toe leaving the ground. There are reports that optimal foot propulsion can be achieved with initial heel contact and termination in the toe leaving the ground (as in normal gait) when the seat height is low (Buck, 2004). However, the results of the current study (average speed, stride length, and APGRF) showed lower values when the seat height was set at −60 mm compared with the seat height being set at −40 mm. APGRF affects the propulsion force of the sagittal plane, therefore, speed is reduced with a reduction in APGRF. A reduced stride length is due to the correlation between
Table 1. Spearman’s rank order correlation coefficient values of speed, stride length, cadence, each joint flexion angles at initial contact (FA-IC) and range of motion during the propulsion period (ROM-PP).

<table>
<thead>
<tr>
<th>Seat height setting</th>
<th>Speed</th>
<th>Stride length</th>
<th>Cadence</th>
<th>Trunk FA-IC</th>
<th>Hip FA-IC</th>
<th>Knee FA-IC</th>
<th>Ankle FA-IC</th>
<th>Trunk ROM-PP</th>
<th>Hip ROM-PP</th>
<th>Knee ROM-PP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Speed</td>
<td>0.73**</td>
<td>0.86**</td>
<td>-0.15</td>
<td>-0.49*</td>
<td>-0.66*</td>
<td>0.24*</td>
<td>-0.26*</td>
<td>0.28</td>
<td>-0.19</td>
<td>-0.17</td>
</tr>
<tr>
<td>Stride length</td>
<td>0.89**</td>
<td>-0.43</td>
<td>0.43</td>
<td>0.17</td>
<td>0.66*</td>
<td>0.41**</td>
<td>0.32**</td>
<td>0.11</td>
<td>0.27*</td>
<td>0.30</td>
</tr>
<tr>
<td>Cadence</td>
<td>-0.15</td>
<td>0.43</td>
<td>0.04</td>
<td>-0.78*</td>
<td>-0.65**</td>
<td>0.43</td>
<td>-0.28</td>
<td>0.17</td>
<td>0.16</td>
<td>0.31</td>
</tr>
<tr>
<td>Trunk FA-IC</td>
<td>0.35**</td>
<td>0.49*</td>
<td>-0.49*</td>
<td>-0.78*</td>
<td>0.31</td>
<td>0.01</td>
<td>-0.26*</td>
<td>0.28</td>
<td>-0.19</td>
<td>-0.17</td>
</tr>
<tr>
<td>Hip FA-IC</td>
<td>-0.22</td>
<td>-0.42</td>
<td>-0.65**</td>
<td>-0.78*</td>
<td>0.31</td>
<td>0.01</td>
<td>-0.26*</td>
<td>0.28</td>
<td>-0.19</td>
<td>-0.17</td>
</tr>
<tr>
<td>Knee FA-IC</td>
<td>-0.56**</td>
<td>-0.35</td>
<td>0.30</td>
<td>-0.28</td>
<td>0.28</td>
<td>0.03</td>
<td>-0.04</td>
<td>0.11</td>
<td>0.27*</td>
<td>0.30</td>
</tr>
<tr>
<td>Ankle FA-IC</td>
<td>-0.62**</td>
<td>-0.50*</td>
<td>0.78**</td>
<td>0.01</td>
<td>0.28</td>
<td>0.03</td>
<td>-0.04</td>
<td>0.27*</td>
<td>0.30</td>
<td>0.19</td>
</tr>
<tr>
<td>Trunk ROM-PP</td>
<td>0.29**</td>
<td>0.20</td>
<td>0.21</td>
<td>-0.16</td>
<td>-0.28*</td>
<td>-0.19</td>
<td>-0.17</td>
<td>0.78*</td>
<td>0.17</td>
<td>0.16</td>
</tr>
<tr>
<td>Hip ROM-PP</td>
<td>0.75**</td>
<td>0.59**</td>
<td>0.62*</td>
<td>0.31</td>
<td>-0.66*</td>
<td>-0.19</td>
<td>-0.17</td>
<td>0.28</td>
<td>0.17</td>
<td>0.78*</td>
</tr>
<tr>
<td>Knee ROM-PP</td>
<td>-0.13</td>
<td>-0.02</td>
<td>-0.30</td>
<td>0.78*</td>
<td>0.32**</td>
<td>0.17</td>
<td>0.16</td>
<td>0.27*</td>
<td>0.30</td>
<td>0.19</td>
</tr>
<tr>
<td>Ankle ROM-PP</td>
<td>0.19</td>
<td>0.31</td>
<td>0.16</td>
<td>0.14</td>
<td>-0.33**</td>
<td>0.12</td>
<td>0.25</td>
<td>0.27*</td>
<td>0.30</td>
<td>0.19</td>
</tr>
</tbody>
</table>

*p < 0.05; **p < 0.01.

4One-way ANOVA, effect of seat height setting (p < 0.01).

Fig. 4. The average values for a speed, a stride length, a cadence, a each joint flexion angles at initial contact (FA-IC) and the range of motion during the propulsion period (ROM-PP) for each seat-height settings are present. *One-way ANOVA, effect of seat height setting (p < 0.01).

speed and stride length. The results of these studies suggest that optimal foot propulsion cannot be achieved when the seat height is set too low.

**Speed, Stride Length, and Cadence**

The results show that stride lengths were smaller than average gait stride length of healthy individuals (1.4 m) for all seat-height settings; and slower speeds than the average gait speed of healthy individuals (84 m/min; Kirsten, 2006) when the seat-height setting was set at +60 mm in relation to leg length (Kirsten, 2006; Whittle, 2001). However, other settings gave faster results than 84 m/min. A strong correlation between stride length and speed was observed. In previous studies, gait speed was reported to be affected by cadence more than stride length (Tanaka et al., 1981). Even in cases where the stance becomes shorter due to differences in seat-height settings, the average speed appeared to increase compared with gait, since stride length increases with inertia while the wheelchair is moving.

**Each Joint FA-IC and ROM-PP**

In cases where the seat height setting was set higher than ±0 mm, no differences in hip FA-IC based on differences in seat height setting were observed; a difference was observed in ankle FA-IC. Extension of the hip joint was restricted by the wheelchair seat and cushion, and it became difficult for participants to make initial contact with the ground using their heel. Frictional force could be amassed and propulsion carried out through ankle plantar flexion and grounding the toes on the floor. In cases where the seat height setting was set at ±0 mm lower, no differences in knee and ankle FA-IC based on the difference in seat height setting were observed; but a difference in hip FA-IC was observed. It is therefore suggested that the floor grounding position of the heel is adjusted by hip FA-IC, not knee FA-IC.

**GRF**

When a wheelchair is in the sitting position, roughly 33% of the weight is supported by the wheelchair’s seat, back support and arm supports (Gutierrez et al., 2006). For wheelchair users, the weight supported by the legs is different from that in normal gait individuals. The VGRF increases as the seat-height setting is lowered. This provides the ability to adjust the load applied to the legs by the seat-height setting, which helps to prevent atrophy in anti-gravity muscles of the legs (Edgerton et al., 1995; Ohira et al., 1992). However, in the current study, APGRF, cadence, stride length and speed declined when the seat height was set at −60 mm. This suggests that the wheelchair was harder to propel when the seat height was set below −40 mm.
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Conclusion

Speed, hip ROM-PP, VGRF, and APGRF increase as the seat height was set lower in this study; however, this effect diminished when the seat-height was set below −40 mm. GRF increased as the seat height was lowered. The results suggest that the effect of the seat-height can be attributed to hip ROM-PP and that the wheelchair is harder to propel when the seat height is set below −40 mm. The physical functions of individual wheelchair users (e.g., foot or trunk function) are different. Therefore, it is necessary to consider not only dynamic seating, but also environment and static seating of wheelchair users. Foot propulsion of the wheelchair can be achieved by setting the seat height according to a combination of physical characteristics, such as the user’s physical functions, leg muscles, and range of motion.

This study contains the following limitations. The sample size was small. Participants were not actual wheelchair users; they therefore had fundamentally different propulsion characteristics. It was also impossible to calculate the moment of each joint because the weight was supported by the wheelchair. The only adjustment made to wheelchairs was to seat height. Seat angles and back support angles may also have important effects on each joint’s range of motion and GRF.

Future wheelchair propulsion analysis will look at seat and back support angles. We aim to test these angles using healthy individuals to attain highly reproducible data, and gain information that can be used for the prevention of secondary disabilities resulting from long-term wheelchair use.

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