‘Every Push Matters’

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I, Lynne Hills, confirm that the work presented in this thesis is my own. Where information has been derived from other sources, I confirm that this has been indicated in the thesis.

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Abstract

Rear axle adjustment has an effect on the stability of a userõs wheelchair. On delivery, a wheelchairõs axle is usually set in its most rearward and most stable position, with guidelines and cautionary advice on its forward adjustment. This is contrary to current clinical recommendations, which advise practitioners to; óadjust the rear axle as far forwards as possible without compromising the stability of the userô (Paralyzed Veterans of America 2005). Thus, clinicians adjust the rear axle forward incrementally, working with the wheelchair user, in order to maintain safety and maximise performance. Theoretically, a more forward axle position has been shown to decrease rolling resistance by reducing the weight transferred through the front castors (Brubaker 1986). Therefore, most clinicians assume that moving the rear axle forward will make the wheelchair significantly easier to propel.

This study was undertaken to investigate if this is true in straight line pushing tasks; propulsion on lino, propulsion on artificial turf (Astro), ascending a 1:12 ramp and ascending a 3ò kerb[1]. Following rear axle adjustment from the most stable position to the least stable position, castor and pushrim forces were recorded during each propulsion cycle. Tasks were performed by a group of eight experienced manual wheelchair users, all of whom had a spinal cord injury below the level of T1.

To assist in the clinical application of the data a óPerformance Capacity Ratioô developed by Nicholson and colleagues (Nicholson,G et al. 2006), was used. This investigated the relationship between a personõs functional performance and their capacity to perform mobility

[1] These straight line mobility tasks have been referred to throughout the text as ófunctional mobility tasksô

tasks when the Rear Axle Position (RAP) was adjusted. This was expressed as a percentage, to gauge whether a person exceeds their ócomfort zoneô when performing different pushing tasks. The ócomfort zoneô was defined as 80% of the maximum voluntary push force a person was capable of. The study has shown that RAP does affect capacity to perform and that subjects were more likely to exceed their comfort zone when performing tasks in a more stable set up. It concludes that terrain impacts on capacity to perform, as wheelchair users are more likely to reach and exceed their capacity on terrain which imposes the greatest resistance.

The synchronisation of the pushrim and castor force measurements allowed a detailed examination of how the forces changed during a typical propulsion stroke, and how this related to castor loading. It was found that castor loading was significantly affected by the Rear Axle Position (RAP), but this did not translate directly into differences in propulsion forces required to overcome increased rolling resistance for all tasks, except the kerb.

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1 Introduction

The UK built environment is extremely challenging for manual (self propelling) wheelchair users. About 3-4% of (manual) wheelchair users have a Spinal Cord Injury (SCI) with the majority of these sustaining their injuries between 16 and 30 years old (Stover et al. 1995). As a result, they will be using a wheelchair for many years and careful wheelchair selection is therefore critical in promoting the highest level of functional independence and in preventing overuse injuries to the tissue structures of the upper extremities.

Only in the last decade or so have significant numbers of people who had acquired a SCI early in life, survived to what is generally considered old age. One of the problems commonly encountered, in addition to the problems faced by older people generally, is severe pain and loss of function associated with their upper limbs. This is caused by many years of propelling a wheelchair and through the use of their upper limbs for all or most activities required for daily living. Any loss of upper limb function in this population can significantly impact on their performance in their Activities of Daily Living, (ADL) consequently increasing dependency. Between 30% and 75% of manual wheelchair users are reported to have developed shoulder pain during their lifetime (Lal, S 1998; Sie IH et al. 1992; Pentland WE & Twomey LT 1991; Gellmen H et al. 1988). Lal (1998) has shown that, although the majority had not yet produced clinical symptoms, 72% of individuals with an SCI had radiological evidence of degenerative shoulder changes 10 years after injury.

There have been a number of previous studies (Richter et al. 1999; Newsam et al. 1996; Cowan et al. 2008; Kotajarvi et al. 2004; M. L. Boninger et al. 2000; A. M. Koontz et al. 2005; Collinger et al. 2008; J. L

Mercer et al. 2006) that have used an instrumented push rim[2] (SmartWheel) to analyse the biomechanics of wheelchair propulsion during defined mobility tasks. These studies were undertaken to examine whether higher propulsion forces increased the likelihood of long term over-use injuries and to further understand biomechanical principles, including how these can be applied to education and injury prevention, with a goal of achieving greater efficiency and independence.

As a clinician the aim of intervention is to maximise functional potential which will promote independent living. It is recognised that an important factor in succeeding at this is to find the right wheelchair and identify the ‘optimal’[3] wheelchair set-up for the client. Historically, wheelchairs were less adjustable than they are today with many features being fixed and pre-determined by the manufacturer. However, the situation has changed significantly in recent years, as wheelchair design has evolved. This optimal set-up today can include adjustment of the rear axle position, seat and backrest angles and rear wheel camber amongst other things. This thesis focuses on the adjustment of the rear axle position as it is commonly assumed by clinicians that moving the rear axle towards the front of the wheelchair will make the wheelchair significantly easier to push and turn. This is backed up by literature as theoretically this adjustment reduces the weight through the front castors, thus reducing rolling resistance (Paralyzed Veterans of America 2005; Brubaker 1986). To date there have been no studies that have measured this.

However, RAP adjustment affects rearward stability of the wheelchair. This is an important consideration for any wheelchair assessment and prescription process as it also is a contributor to serious injuries or even death of users when too unstable (Calder & R. L. Kirby

[2] Push rim and hand rim are interchanged throughout the text.

[3] ‘Optimal’ is a term widely used in the clinical setting to describe a wheelchair that has been set up to meet an individuals specific seating and performance needs.

1990; Unmat & R. L. Kirby 1994). The National Prosthetic and wheelchair services Report 96/97 states that there are 750,000 wheelchair users in the UK, representing approximately 1.5% of the population. Some 350 of these users are seriously injured because of tipping incidents. The MHRA highlights the majority of instability incidents (51%) are related to rearward stability with 39% relating to forwards stability and only 10% sideways (Medicines and Healthcare Regulatory Advisory 2004).

There has been extensive work carried out on wheelchair static stability (R. L. Kirby et al. 1989; R. L. Kirby 1996; R. L. Kirby & Dupuis 1999; R. L. Kirby et al. 1994; RL Kirby et al. 1995; Calder & R. L. Kirby 1990; R. A. Cooper et al. 1994; Loane & RL Kirby 1985; Tomlinson 2000). Key information that can be drawn from these studies is that they all consider safety to be the paramount concern in the prevention of tipping. However, when working with active and/or experienced wheelchair users, stability has, to some extent, been compromised to achieve peak performance and to assist in the development of more advanced wheelchair skills

A key objective for this study was to measure functional performance over the different terrains found to be typical of everyday wheelchair use. In particular, it sought to examine the effects of a less stable or ĩtippyò wheelchair on the pushing parameters and whether these individuals were more or less likely to exceed their capacity to perform when completing such tasks. The overall goal of the research was to increase awareness of the implications of axle adjustment and, using research based evidence, provide an insight into how a less stable wheelchair performs on everyday terrains.

2 Literature Review

This chapter considers the current literature concerned with the performance of self-propelling manual wheelchair users in relation to the RAP of the wheelchair setup. Although little research has so far been completed on adjustment of the RAP and it impact on a users ability to perform functional mobility tasks, results and theories born out of previous projects in related areas are available. The literature review comprises of a brief analysis of work complete to date, key findings and reference to national and international studies.

Section 2.1 discusses the assessment, provision and design of wheelchairs; Section 2.2 defines static wheelchair stability; Section 2.3 identifies current measurements and standards for static wheelchair stability; Section 2.4 highlights literature on wheelchair features and their impact on wheelchair stability; in Section 2.5 rolling resistance and its influence on stability is discussed; Section 2.6 challenges the standards and discusses the balance between stability and function; Section 2.7 introduces wheelchair propulsion biomechanics; Section 2.8 explores the influence axle position has on wheelchair biomechanics leading onto Section 2.9 which explores key aspects of kinematic measurement in wheelchair performance. The literature review concludes in Section 2.10 which explores clinical application of data generated.

2.1 Wheelchair - Assessment and Provision & Design

In the UK, NHS wheelchair provision is delegated to a network of Regional Wheelchair Services, which remain responsible for the prescription and provision of all types of wheelchair. These services include those needed for people who use wheelchairs episodically, as well as those who require use of a wheelchair to meet all of their mobility needs. The latter is known to include those who have a spinal cord injury, a group representing only a small percentage of total service users (Lachmann et al. 1995). However, due to their specific wheelchair requirements, they are amongst the most demanding in terms of their functional goals and long term needs.

Clinical experience has shown that the services offered and the type of wheelchairs available for this particular client group varies considerably between regional services. There is an increasing demand in the UK for lightweight, multi adjustable wheelchairs for such users, which consequently creates further strain on the already stretched resources of the wheelchair services. This makes it extremely difficult to meet the needs of their clients (Rose & Ferguson-Pell 2002).

This was also seen in a comprehensive review carried out by the Audit Commission and outlined in the report óFully Equippedô (Audit Commission 2000). The report demonstrated that there is an inequality in provision, despite the setting of good service standards by the Department of Health. Some wheelchair services do provide a wide range of wheelchairs and equipment, whereas others have a tight eligibility criterion which limits such specific provision. Such policy can be related to the funding that each service has available to them and the needs of the local population (e.g. children, adults etc). This report also identified that access may also be greatly influenced by those who conducts the assessments. Some services have become heavily dependent on commercial suppliers for clinical expertise. The therapistôs role should be to specify the clinical context and goals that the equipment should address, along with how the wheelchair should be configured/set-up. Suppliers have product specific skills that are employed to recommend the products that meet a certain need. It should remain the therapistôs role to make the ultimate buying decision and to ensure the wheelchair is optimally set-up for performance, a view supported by the Fully Equipped report (Audit Commission 2000).

Pope (2005) suggests that one possible reason for deviation from this model of service provision is that clinicians often lack the skills and access to training needed to work in this field. There is no standard requirement for clinicians working within the field, and thus there is considerable scope for inappropriate prescriptions to be made (Pope 2005). Such practices can be detrimental to both the user and the service, supporting the researcher's view that there is very little evidence based practice filtering through to a clinical level.

In their first study looking at discharge of the Spinal Cord Injured patient, Rose & Ferguson-Pell (2002) reported that 46% of users changed their wheelchair, provided by the Wheelchair Service, within the first year of discharge and that the chairs they were prescribed were minimally adjustable. However, in their recent repeat study looking at the same population, it showed that users were more satisfied with their provision and less likely to change their wheelchair. There was however, an increase in users being prescribed more adjustable lightweight wheelchairs and also in those accessing the voucher scheme and alternative funding resources, reflecting more choice in the range of wheelchairs available to them (Rose & Ferguson-Pell 2009).

Typically, a person with a spinal cord injury tends to be a young male, usually healthy with normal life expectancy and with the potential to lead an active life (Rose & Ferguson-Pell 2002). Their lifestyle may include work and leisure pursuits and often a return to driving, including lifting the wheelchair in and out of a car. Key to restoring a fulfilling lifestyle is effective mobility within a wide range of environments, and effective mobility is achieved through effective wheelchair provision and set up.

2.2 Wheelchair Stability

There has been much research into the testing of wheelchair stability in a static state (R. L. Kirby et al. 1989; R. L. Kirby 1996; R. L. Kirby & Dupuis 1999; R. L. Kirby et al. 1994; RL Kirby et al. 1995; Calder & R. L. Kirby 1990; R. A. Cooper et al. 1994; Loane & RL Kirby 1985;

Tomlinson 2000). Stability comprises a number of different aspects, including rearward, forward and lateral stability. The determination of each of these is important, but the focus of this study is primarily on rearward stability. Causes of rearward instability include, leaning backwards, acceleration forwards, ascending kerbs and slopes. An American study found that 77% of wheelchair deaths were related to rearward instability incidents (Calder & R. L. Kirby 1990).

This study focuses on rearward stability specifically not only because it is the cause for the majority of tipping accidents but because it is affected by adjustment of the Rear Axle Position (RAP) which in this thesis is key.

2.3 Measurement of Rearward Stability

Majaess et al (1993) describes static rearward stability as the angle away from the horizontal surface where a tipped wheelchair is critically balanced. This provides a ótipping angleô A larger angle indicates increased stability. This is determined by the position of the óCentre of Massô (COM) of the system in relation to the axis of rotation (Majaess et al. 1993; Tomlinson 2000).

Static stability testing is carried out using a tilted platform. The high reliability of this test method has been well documented (Loane & RL Kirby 1985; R. L. Kirby et al. 1989; R. L. Kirby et al. 1996a). Testing with such a platform is usually carried out using anthropomorphic test dummies, until the wheelchair and the dummy are tilted to or past the point of instability, as demonstrated in Figure 1

Figure 1 - Static stability testing platform (Rentschler 2002)

A DHSS Technical Bulletin (TB/SA/6) set standards of 12° (manual) and 16° (power) as a safe angle of stability. However, the MHRA have since set guidelines that do not reference these standards and advise instead that referral should be made to manufacture guidelines (West Midlands Rehabilitation Centre 2005). There appears to be a lack of understanding around wheelchair stability, giving rise to difficulties equating the usage of angles, stated in wheelchair manufacturing guidelines, into practical terms.

The MHRA (2004) booklet titled, ‘Guidance on the Stability of Wheelchairs’, was devised as a result of concerns around users unwittingly altering their wheelchair stability through mechanical means (RAP adjustment), thus increasing the risks taken (Medicines and Healthcare Regulatory Advisory 2004). The booklet sets out all the elements that can affect stability, again identifying that reference should be made to manufacturer’s instructions on wheelchair stability prior to attempting any functional tasks. What the booklet does not refer to are individual abilities or skills in management of a more unstable wheelchair, nor does it discuss the differences in propelling a stable versus an unstable wheelchair. There needs to be some recognition that that there is often a compromise between stability and function.

With 12° (manual) and 16° (power) degrees accepted as the recommended guide, it does not provide sufficient information for looking at function, or enough specific data to carry out risk assessment in any specific individualôs environment.

2.4 Wheelchair features influencing stability

A óstandard wheelchairô, those models most commonly issued by wheelchair services, is typically more stable than a lightweight or multi-adjustable chair, or those defined as óhigh performingô or óactive userô chair (Loane & RL Kirby 1985). It is the lightweight, multi-adjustable wheelchair that is usually prescribed to those with a spinal cord injury, as they are designed to offer adjustability of the various components in order to achieve optimal set-up. However, such adjustability of these wheelchair components also allows them to be configured into a potentially óunstableô chair.

Wheelchair features and their effects on stability have been well documented (Trudel et al. 1997; Majaess et al. 1993; R. L. Kirby et al. 1994; Brubaker 1986; M. L. Boninger et al. 2000). The features need to be fully considered when prescribing any wheelchair and determining the stability of a chair. Such features include; **Castors:** (R. L. Kirby et al. 1994); **Camber:** (Trudel et al. 1997) **Wheelbase:** (Majaess et al. 1993; Tomlinson 2000) **Brakes:** (Loane & RL Kirby 1985) / (Trudel et al. 1997)/ (R. L. Kirby & Dupuis 1999); **Anti tips** (R. L. Kirby et al. 1994) and **Adding loads** (R. L. Kirby et al. 1996b)**.** The feature of particular importance in relation to this study involves the **Axle.**

By moving the rear axle forward (

Figure 2), rearward stability is decreased (Majaess et al. 1993). Wheelchairs are usually delivered with the axle set in its most rearward position, with guidelines and cautionary advice on its adjustment. As previously described, neither guidelines for setting up the rear axle nor advice to clinicians on best-prescribing practice are included on delivery of the wheelchair, a view also supported by Tomlinson (Tomlinson 2000)). Brubaker (1986) recommends that the axle should be moved forwards incrementally, provided that the wheelchair user feels stable. A key element in using this approach is feedback from the user about how stable they feel, which must remain a priority.

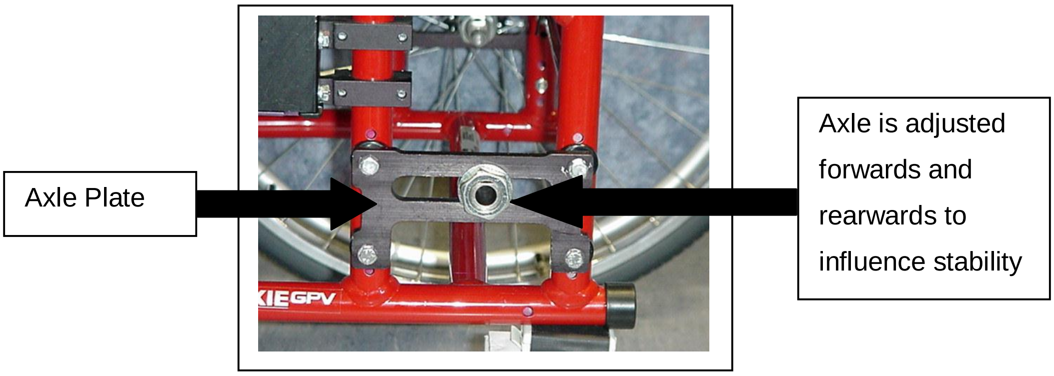

Figure 2 - Adjustable axle on a GPV lightweight wheelchair

Adjustment of the horizontal position of the rear wheel maybe one of the most important adjustments to a manual wheelchair, not only for its safety implications but its direct affect the rolling resistance, ease of propulsion, required stroke frequency, hand contact angle and stability (Brubaker 1986; M. L. Boninger et al. 2000; DiGiovine 2006), and thus impacts directly on the user’s abilities. If this does remain such an important influence on propulsion, one must question as to why more established guidelines to assist clinicians with set-up do not exist.

A change in the horizontal position of the rear axle position changes the location of the wheels with respect to the user’s centre of gravity. The rear wheel has a lower rolling resistance than the castor (due to the radius of the wheels), the materials used and the distribution of weight between the wheels. Rolling resistance can best be described as the force to be overcome by the user to keep the wheelchair moving at a constant velocity over a particular surface (Tomlinson 2000). The rear wheel has the lowest rolling resistance since its radius is much greater than the castor. It is therefore important that the rear wheel is located forwards, as close as possible to the user€s centre of gravity, reducing the overall rolling resistance of the wheelchair. To reduce the likelihood of upper limb injuries through the repetitive motion of upper extremity manual wheelchair propulsion, the most appropriate set up is believed to be shifting the axle forwards and upwards (Tomlinson 2000) .

Kauzlarich and Collins (1988) highlight that excess rear stability can limit the ability of a wheelchair user to; lift the front wheels and to perform functional tasks, increase a wheelchairs rolling resistance, decrease traction and increase downhill-turning tendency on side slopes. Faced with such consequences, it is perhaps not surprising that many active wheelchair users opt for a wheelchair with minimal stability.

2.5 Rolling Resistance

As already highlighted, a change in a wheelchair€s RAP contributes to the proportion of force being transferred through the castors (Brubaker 1986). The percentage of force going through the castors is increased when the rear axle position is moved rearwards (Tomlinson 2000; Paralyzed Veterans of America 2005). This in turn is thought to increase the rolling resistance of the wheelchair (Brubaker 1986).

Many factors influence the rolling resistance of self-propelled wheelchairs. Among the influences are tyre characteristics, such as size, tread, rigidity and inflation (Tomlinson 2000; Brubaker 1986), Spinal Cord Injury Peer Support 2008). Rolling surfaces such as tile, pavement, carpet or stone also affect the amount of resistance. The latter, however, are extrinsic factors that often cannot be altered, particularly outdoors. Indoor surfaces such as carpet can be modified more easily, based on costs and needs of the wheelchair user.

However, although influenced by the surface the wheelchair user is traversing, it is also affected by the mass distribution on the wheels, wheel radius, total mass and specific tyre characteristics (Zatsiorsky 2000; B J Sawatzky et al. 2005). Whilst studies do not address the specific position of the wheelchair axle directly, there is much discussion about wheel placement in relation to the amount of effort required by the user.

Tomlinson (2000) states that ñtotal rolling resistance is reduced as a larger proportion of weight is redistributed to the rear wheelsò (p904). As the rear wheels are attached directly to the rear axle, its placement on the wheelchair becomes an important factor in rolling resistance. Tomlinson (2000) further states that the wheel's rolling resistance is inversely proportional to its radius, so the rolling resistance coefficient is smaller for the rear wheels than the castors (small front wheels). This factor stresses the importance of a wheelchair design/adjustment that places weight distribution over the rear wheels and axle.

Consulting engineers for Spinal Cord Injury Peer Support (2008) also stress the importance of rear axle and rear wheel placement on the wheelchair. Again, a position in which the userõs centre of mass is positioned directly above the rear axle is recommended. They recommend that in the vertical direction, the user should be positioned so that their fingertips can touch the rear wheel axle, a view supported by other studies (L H van der Woude et al. 1989; Paralyzed Veterans of America 2005; Nicholson,G et al. 2006) . These studies believe that, not only does the position of the userõs weight above the rear axle reduce friction, it allows for more power to be transmitted to the hand rim.

Buning & Schmeiler explain that reaching as far back as possible on the hand rim gives the user a pushing stroke that has two parts: flexion and extension. They also explain that this is achieved when the wheelchair userõs shoulder is in alignment with the rear axle (Burning &

Schmeiler 1999). A common focus of rolling resistance in self-propelled wheelchairs is positioning with respect to the rear axle. While some experts focus on friction coefficient, others focus on the physical wear and tear on the user, as more efficient wheelchair use reduces injury and strain (M. L. Boninger et al. 2000; L H van der Woude et al. 1989; De Groot et al. 2002; van Drongelen et al. 2005; J. L. Mercer et al. 2006; Collinger et al. 2008). However, there has been no development of a practical clinical tool that can help clinicians in everyday practice.

2.6 Stability & Functional Performance: Functional or safe?

Kirby & Dupuis identified optimal angles for set up of **12.3º** with rear wheels locked (brakes applied), and **20.2º** with the rear wheels unlocked (brakes removed) (R. L. Kirby & Dupuis 1999). They recommended that their findings could be used as a rule of thumb clinically, knowing that individual users would vary. However, they should not be prescribed as a standard of practice.

This presents the active wheelchair user with a number of different problems. An óactive wheelchair userôcould probably be best described as someone who carries out the following activities; ascending and descending kerbs and slopes; performs back wheel balances to carry out wheelchair skills; completes numerous functional transfers daily; accesses public transport. Should an active wheelchair user be set up with a tipping angle of 12°, it is likely that they would find performing functional tasks extremely difficult, as the amount of push rim force required to perform such tasks would be greatly increased in such a stable set up. For example, experienced wheelchair users, as part of the WOWSUP (2006) study, were performing in wheelchairs set up with a tip angle that was as small as 4.5 degrees (Nicholson,G et al. 2006), compared to the standard guidelines set of 12 degrees (Medicines and Healthcare Regulatory Advisory 2004). This, arguably, is a risk which needs to be managed to help promote propulsion performance and perhaps reduce risks of overuse injury, which may occur if higher rolling resistance forces have to be overcome when the wheelchair RAP is in a more stable position. It could be argued that clinicians need to take responsibility for conducting clinical risk assessments where appropriate and that such inflexible guidelines should not dictate clinical practice.

Kirby stated that stability is modifiable (R. L. Kirby 1996). Wheelchairs can be adjusted in many ways to achieve the optimal performance. Optimisation of setup is a lifelong and evolving process, it is dependent on how the wheelchair user adjusts their skill development, as well as their changing levels of health and ability. It is essential that clinicians assist effectively in the optimisation process, by identifying where to start with axle positioning and in allowing for the wheelchair user to progress their skills and performance of functional mobility tasks. It is therefore felt to be important, when looking at functional performance, that propulsion biomechanics and its effect on performing mobility tasks are explored.

2.7 Wheelchair Propulsion Biomechanics

With a gross mechanical efficiency of around 10%, wheelchair propulsion, as a mode of ambulation, is inefficient, showing that wheelchair users operate much higher levels of energy expenditure, force and power to achieve independence in mobility (De Groot et al. 2002). This is partly due to the relatively small muscle mass of the upper limbs, biomechanical difficulties involved with the "coupling / decoupling of the hand to the rim", trunk movement and a large recovery phase (De Groot et al. 2002). Over time, it is thought that this produces overuse syndromes, injury and pain (Kotajarvi et al. 2004; J.L Mercer et al. 2006). Mercer et al also found that people who experienced larger forces and moments were more likely to have coroco-acromial pathology or to exhibit signs of pathology on physical examination (as cited in (Collinger et al. 2008)).

Repetition is also a major risk factor in developing injuries and one should consider the number of times a wheelchair user must negotiate small steps, slopes, as well as transfer. In a study by Van Drongelen, it is suggested that neither, propulsion nor Activities of Daily Living (ADL) tasks (by themselves), are responsible for the high number of overuse injuries. It is suggested rather, that it is the combination of the two that forms the high risk (van Drongelen et al. 2005). Subberao et al (1994) found that individuals with a spinal cord injury failed to find relief from the majority of treatments available. They believe this was due to the need of the individual to carry out unavoidable tasks during the day (as cited (Rice et al. 2008). This is clearly a subject which would benefit from further research (H. E. J. Veeger et al. 2002) .

Manual wheelchair propulsion and wheelchair sports have increasingly become the subject of detailed biomechanical analysis (Vanlandewijck et al. 2001). Research into wheelchair propulsion has been undertaken to assist in optimising performance and minimising upper extremity loading and this has assisted in the development of clinical guidelines developed by the Consortium for Spinal Cord Injury (Paralyzed Veterans of America 2005). It is important to study the causes and consequences of these high loads on the upper extremity, as well as study the activity of wheelchair propulsion over different terrains. Recent studies have examined kinematic data using motion capture sensors (J.L Mercer et al. 2006; Collinger et al. 2008) and ultrasound (Brose et al. 2008) to further understand the demands placed on manual wheelchair users. All of these conclude that body weight was a variable for affecting shoulder forces and that users who did experience shoulder pain, did not necessarily develop higher propulsion forces. This suggests that propulsion biomechanics contribute to pathology, rather than pathology influencing propulsion style (Collinger et al. 2008). From the literature, there is a general consensus that we have to further develop our knowledge in order to establish how we can assist long term propellers to minimising their risk of upper limb injury.

2.8 Propulsion Biomechanics and Axle Position

RAP has been shown to influence the smoothness and the frequency of pushing a wheelchair. This is demonstrated in a study by Masse (1992) who examined propulsion forces and their relationship with the rear axle position, in five paraplegic subjects, in six different seating positions (Masse et al. 1992). The kinematic analysis revealed that a joint range of movement in the upper limbs was smoother for the lower and forwards position. This more forward position of the RAP was also supported by Boninger (M. L. Boninger et al. 2000). He found that wheelchair users with their RAP set in a forward position, would have reduced propulsion frequency and spend more time on the pushrim, with consequently lower push rim forces. However, there is a contradiction between the two studies; Boninger supports a high axle position rather that the low position advocated by Masse (1992). Both studies conclude that the more forward the axle, the better the propulsion biomechanics.

Previous studies have hypothesised that decreasing the frequency of propulsion may help prevent median nerve injury (M. L. Boninger et al. 1999; M. L. Boninger et al. 2000; Leibel & Patrick 1998). Frequency of propulsion is decreased when the user can access or reach the hand rim on the upstroke, as well as the forward motion (finger tip to axle usually acts a guide for achieving this (Paralyzed Veterans of America 2005). Muscle use is redistributed more evenly throughout the upper arms, thereby reducing overuse injury and fatigue. The user also generated much more power with each arm movement (Leibel & Patrick 1998). Therefore, if push frequency is indeed reduced with the rear wheel positioned forwards, this would appear to assist in the prevention of overuse injury.

This was supported by (Richter 2001), who used a force sensitive push rim called a SmartWheel to examine wheelchair propulsion biomechanics of five wheelchair users. The research studied seat position on hand rim moments, joint kinematics, joint torques, push frequency and push angle using a quasi-static wheelchair propulsion model. The results found that decreasing the distance between the shoulder and the axle position increased push angle and elbow extension torque, whilst decreasing push frequency and shoulder torque.

However, when Rasmussen et al (2004) explored ergonomics of wheelchair propulsion and the relationship between push angle and Gleno-humeral (GH) forces, they identified that whilst a more forward axle did lead to smaller GH forces, if the push angle progressed to 30 degrees, the initial force to the GH joint became higher. With the axle positioned to the rear, they observed improved access to the wheel. This allowed the hand to push the rim in a more vertical direction, consequently allowing good GH articulation. They advised that wheelchair users should push downwards instead of forwards to keep GH forces to a minimum.

The position of the rear axle is a key set-up parameter when prescribing wheelchairs. Studies of rear axle position and propulsion have focused on energy cost and respiration in relation to seat position (Mijis et al 1989, van de Woude et al 1990) upper limb kinematics in relation to rear wheel positioning (Masse et al 1992) the effects of seat height on push frequency and torque (Richter 2001) and rear wheel effect on comfort, push frequency and stroke angle (Samuelsson et al 2004). These studies explore the height adjustment of the rear axle position, but not its adjustment in the forward (less stable or ‘tippy’[4]) and backward (more stable) positions. However, no other known studies have examined

4 The term ‘tippy’ means ‘less stable’ and is used throughout the thesis as it is a word commonly used clinically. It is a more neutral a statement than ‘stable’ or ‘unstable’ which have a connotation specifically in relation to safety.

the effect of the rear wheel axle position and its impact on performing functional mobility tasks in a more or less stable axle position.

2.9 Biomechanics and functional performance

To achieve independence, wheelchair users must have the ability to negotiate everyday obstacles such as ramps, door thresholds, uneven terrain and carpets. Such challenges require skill, effort and determination. A wheelchair should be seen as a tool that can be adjusted and configured to an individualôs needs, with a view to limiting the effort required to perform such obstacles. Clinical guidelines for the preservation of upper limb function following a Spinal Cord Injury have been developed by the Paralysed Veterans of America (Paralyzed Veterans of America 2005). These guidelines recommend the minimisation of push force and frequency of repetitive upper limb tasks and the use of long strokes during propulsion.

In order for such measures to be calculated accurately, measurement tools have been developed to assist in the gathering of such data. The SmartWheel is a modified wheel instrumented with a 3-beam system that allows for the determination of 3-dimensional forces and moments (Three Rivers Holdings). The SmartWheel can be mounted on the individualôs own wheelchair therefore wheelchair-user interface and external conditions can be simulated (Vanlandewijck et al. 2001). The SmartWheel contains an on-board optical encoder that determines the rotational angle of the wheel. (A. M. Koontz et al. 2005)

As a result of such development a Smart wheel Users Group was established in 2004. The group was formed as a central hub for all those involved in its usage and it aimed to create a standard clinical protocol, populate a database for the developed protocol and create reference values for specific measures. From this a Smart Wheel Users Guide was developed (Three Rivers Holdings). The SmartWheel is intended to facilitate the development of normative standards. Development of such standards was aimed at allowing the identification of those individuals for whom manual wheelchairs are an appropriate prescription and providing information on the effect different wheelchair setups has on performance (Three Rivers Holdings).

Subsequently, the Smart Wheel users group identified four parameters generated by use of a Smart Wheel that would be the most clinically relevant when attempting to preserve upper limb function. These are outlined in Cowan's study (Cowan et al. 2008). The SmartWheel is a measurement device that attaches to a variety of wheelchairs, used in the clinical setting to measure parameters involved in the movement of the wheelchair. Four key parameters are: velocity (considered in Section 2.9.1), push force (2.9.2), push frequency (2.9.3) and stroke length (2.9.4).

2.9.1 Wheelchair Velocity

Wheelchair velocity can be best described as the speed (metres per second) that the wheelchair is moving in the direction of travel. There has been some discussion towards the minimum and ideal velocity requirements for safe and active wheelchair use. Hoxie and Rubenstein (1994) found that a velocity of 1.06 m/s represents the average minimum velocity needed to safely cross an intersection (Hoxie & Rubenstein 1994). This was chosen as a threshold velocity in a recent study by Cowen (2008). This study found that velocity ranged from 0.8 to 1.6 m/s for propulsion on a level surface (Cowan et al. 2008) and similar results were seen in Kotajarvi et al (2004) and Koontz et al (2005) (Kotajarvi et al. 2004; A. M. Koontz et al. 2005).

Newsam (1996) also conducted a study looking at the effects of terrain on propulsion. The research found that when wheeling over carpet, the velocity of propulsion was significantly slower than the tile condition. It also found that individuals with high level spinal cord injuries had an even slower velocity, suggesting that users with higher lesions must work near or at their maximum capability for basic community functions (Newsam et al. 1996).

With reference to wheel position and its effect on velocity, little research has been reported. In a study by Walsh et al (1986) the relationship between seat position and linear velocity in wheelchair sprinting was investigated. Testing was conducted with nine male subjects with various physical disabilities, pushing at maximum speeds on an ergometer. The results revealed no significant differences between the maximal linear velocities at each of the nine seat positions chosen for investigation. These findings suggest that given a limited variability in seat positions, maximal linear velocity will be minimally affected (Walsh et al. 1986).

2.9.2 Push Force and Moments

While wheelchair users encounter many surfaces in everyday life, studies on wheelchair propulsion biomechanics have typically been carried out on either treadmills or in laboratory settings, away from normal environmental conditions. Koontz et al (2005) supported this view and whilst acknowledging the benefits of laboratory based trials, moved away from such traditional methods onto testing wheelchair performance over different terrains. The study looked at wheelchair motion at start-up (first push of the run), rather than the steady state responses to surface resistance (last three pushes of the run). It identified that the start up phase generated greater propulsion force and torque (Push 1 - 103.2 +/- 24.4N) and Push 2 (101.8 +/- 30.7N) compared to steady state (63.6 +/- 2.9N) on concrete surfaces and that these forces were greater on those surfaces imposing greater resistance (A. M. Koontz et al. 2005). In comparison, Koontzӳ results on tiles show lower forces for push strokes 1 and 2, but higher for the steady state, at (89+/-27.1 N) (Cowan et al. 2008).

DiGiovine (1997), used the SmartWheel for determination of propulsion forces and the amount of work required to propel over a series of terrains, including a bumpy tile, a sloped tile (1% grade), a flat tile and carpet at slow, medium, and fast self-determined speeds (DiGiovine et al. 1997) . By contrast to the work of (A. M. Koontz et al. 2005), this study considered only steady-state strokes in the analysis (the last three pushes of the cycle). Although the amount of work required was significantly different between surfaces, the results were formed from a single subject, without a disability, pushing at varying speeds.

2.9.3 Push Frequency

Push frequency can otherwise be described as cadence, or the number of strokes over a given length of time. Cowan et al. (2008) measured cadence at 0.8 to 1.2 cycles per second, for a variety of velocities (Cowan et al. 2008). It is found that moving the rear axle forward several inches, in alignment with the shoulder, reduces cadence and forces on the push rim (M. L. Boninger et al. 2000). Koontz et al's results are similar but slightly higher to those of Cowen, with a cadence of 1.23 +/- .22 cycles per second (A. M. Koontz et al. 2005; Cowan et al. 2008).

Cowen (2008) also indicates that users vary their cadence, depending upon the surfaces they were tested on (Cowan et al. 2008). For example, the cadence of users decreases on surfaces creating less resistance, such as tile. Users increase cadence with more resistant surfaces like concrete or carpet. Cowan's results show that users are able to exert more or less force and change the number of strokes, based on varied surfaces. The result of Boninger's study found that horizontal axle position is correlated with the frequency of propulsion, thus suggesting that the position of the rear axle, in relation to the seat position of the user, will affect cadence (M. L. Boninger et al. 2000).

2.9.4 Stroke Length

The results of cadence, recorded as cycles per second, obtained in Koontz's study were measured with a stroke length of 77.03 +/- 10.21 (A. M. Koontz et al. 2005). However, Cowan's participants were found to select a slower velocity, lower cadence and longer stroke length indicating that a longer stroke length is desirable, as it reduces cadence, thereby reducing repetition of specific muscle groups of users (Cowan et al. 2008). Longer stroke length then arguably might be said to help reduce overuse injuries and muscle fatigue.

2.10 Clinical Application

Clinical meaningful changes have yet to be established for wheelchair propulsion parameters. It could be argued that in order to better understand quantitative measures of wheelchair propulsion and apply this knowledge clinically, it is important to have a measure. In gait analysis, such gait parameters can be related to a "normative" database of walking patterns. However, as humans are not designed to propel wheelchairs, it is particularly difficult to define a "normal" pushing cycle and therefore, such a database is not known to exist for wheelchair users, although the Smart Wheel Users Group is working towards such data. It is important however, to measure what someone can be expected to achieve comfortably, and safely, when performing their daily functional mobility tasks. This is supported by the WHO International Classification of Functioning, Disability and Health 2001 (ICF)[5]. ICF identifies the term "performance" as equivalent to "demand". ICF recommends that individuals should be able to function near to their capacity in their environment; but that it is better that performance should be less than an individual's capacity, in order to avoid injury.

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5 The ICF guidelines can be accessed and explained in more detail at: http://www.who.int/classifications/icf/en/

2.10.1 Performance: Capacity Ratio

There are currently no tools available to assist clinicians in achieving the optimal set up for each individual wheelchair user. WOWSUP (2006) was a project that explored wheelchair optimisation and performance (Nicholson,G et al, 2006). WOWSUP stands for The Workshop for Optimisation of Wheelchair Selection and User Performance. The project introduced an outcome measure relating to the propulsion forces needed to perform an actual task and the capacity of the user to perform it. Functional performance was defined as óthe effort required to push the wheelchair within the userõs own environmentô and their capacity was described as ówhat the user can comfortably achieve with effort when pushing a wheelchairô (Nicholson,G et al, 2006). This is expressed as a percentage and termed the óPerformance Capacity (P:C) Ratioô óCapacityô was defined as the maximum/peak force that each individual applies to the rim in an isometric test (e.g. by pushing as hard as possible on the push rim with the wheels blocked) and the óperformanceô was the force generated on the push rim while performing functional mobility tasks. The P:C Ratio can then be used to gauge whether a person exceeds their available capacity or comfort zone, while carrying out different functional mobility tasks. Through its application there are also parallels seen in the biomechanical parameter ñmaximum voluntary contractionò (MVC) used to normalise muscle contractions. The ratio of a functional muscle contraction to MVC is standard clinical practice and widely used in gait analysis (Barnard. T. 2006).

It is important to note that many manual wheelchair users have very limited upper extremity strength, particularly grip strength. This significantly reduces their capacity to push their wheelchair. However, the demands of the task of pushing are dictated by the demands of the environment. To achieve functional mobility they may have to frequently over-exert themselves and this sets the stage for long term over-use injury. A measure that compares the demands of the task (performance) with the physical capacity of the user, gives a clear clinical indication of the risk of over-exertion of an individual. Clearly without a long-term (15-20 years) study robust evidence for a ñthreshold for over-use injuryò using P:C ratio is not available. However there have been studies relating levels of exertion to maximum voluntary contraction but these are not related to wheelchair propulsion (Barnard. T. 2006).

A wheelchair user may fall well within their available capacity when managing level/flat terrain (a low P:C Ratio) but on ascending a slope may reach, or even exceed, their capacity (a high P:C Ratio). In the WOWSUP study, 80% was chosen by Nicholson et al (2006) as a safe threshold for repetitive wheelchair propulsion forces and suggested that if users exceed this threshold they are óred-liningô using a familiar analogy that is used to protect over-revving of car engines. The 80% threshold for red-lining was based on subjective questioning of subjects in the WOWSUP study. It was found that there was a link between reported levels of exertion and a level of approximately 80% of a subjectõs capacity Nicholson et al (2006).

The WOWSUP study identified that more experienced wheelchair users were less likely to exceed their capacity to perform than inexperienced users. The study also consistently observed the expected higher P:C Ratio when users ascended a slope. The P:C Ratio is also comparable to that of The Borg Scale. This scale is viewed as a simple method of Rating Perceived Exertion (RPE) and is used by coaches to gauge an athlete's level of intensity in training and competition. One of the most common applications of this is the 15 point scale outlined below in Table 1.

15 Point Scale

Point	% Effort	Description
6	20	
7	30	Very, very light (Rest)
8	40	
9	50	Very light gentle walking
10	55	
11	60	Fairly light
12	65	
13	70	Somewhat hard steady pace
14	75	
15	80	Hard
16	85	
17	90	Very hard
18	95	
19	100	Very, very hard
20		Exhaustion

Table 1:The Borg 15 point scale (Borg Scale 2009)

In applying this to the P:C Ratio the 80% value chosen as the threshold for an individual exceeding their capacity can be correlated with the perceived effort of óhardô

This P:C Ratio will be adopted within the current study to explore whether wheelchair users are more likely to reach, or exceed, their capacities to perform with the rear wheel axle in a more or less stable position (Nicholson,G et al. 2006).

It is a challenge for all clinicians working with any wheelchair user to achieve an óoptimalô wheelchair set up. It is essential that key research information is channelled into the workplace and communicated clearly to clinical staff, something which is not seen enough in practice. The main aim of this study is to look at RAP in relation to propulsion forces and identify ónuggetsô of information that can be simply transferred into everyday clinical practice and that it is easily understood. These nuggets will aim to be centred around key influences that the rear wheel axle position has on manual wheelchair use and these will be generated through addressing the defined hypothesis set in the next chapter.

3 Methodology

This chapter of the thesis explores the hypotheses generated for the study and proposes the outline of the study and what measures are involved.

Section 3.1 outlines the hypotheses to be explored; Section 3.2 discusses the process for Ethical approval; Section 3.3 identifies the key measures used in generating the data for the study; in Section 3.4 subject details are presented; 3.5 explains the experimental procedure followed for each subject; Section 3.6 examines the data collection process and analysis methods used; Section 3.7 presents statistical methods employed.

3.1 Hypothesis and Experimental Design

This study investigates the relationship of castor forces and hand rim forces generated dynamically during the propulsion of a wheelchair, when the rear axle is positioned in its most stable and least stable position. These hypotheses are derived from gaps found in the literature and also from a thirst for clinical knowledge in this area to assist in evidence based clinical practice.

3.1.1 Hypothesis 1

H0: There is no significant difference in the castor forces generated during straight line functional mobility task when the rear axle is moved forwards (tippy) compared to the most stable (rearmost) position.

H1: There are significantly lower castor forces generated during straight line functional mobility tasks when the rear axle is moved forward (tippy) compared to the most stable (rearmost) position.

3.1.2 Hypothesis 2

H0: There is no significant difference in the castor forces generated when the rear axle is moved forwards (tippy) during straight line functional mobility over different terrains.

H1: The castor forces are lower when the rear axle is moved forward (tippy) during straight line functional mobility over different terrains.

3.1.3 Hypothesis 3:

H0: There are no significant differences in individual SmartWheel parameters (Peak Mz, velocity, stroke angle and cadence) when the rear axle is moved forward during straight line functional mobility over different terrains.

H1: There are significant differences in individual SmartWheel parameters (Peak Mz, velocity, stroke angle and cadence) when the rear axle is moved forwards during straight line functional mobility over different terrains.

3.2 Ethical Approval

The study took place at the Stanmore Clinical Research Facility (SCRF) at the Royal National Orthopaedic Hospital (RNOH), Stanmore. All participants gave informed consent to participate in the study (see Appendix 1 for all information related to participation and consent).

The proposal for the study was approved by the joint Research and Ethics Committee at RNOH using the National Research Ethics Committee Form and following the COREC approved system.

3.3 Dynamic Stability Measurement

3.3.1 Study Design

A quantitative analytical research design was undertaken to establish forces generated by wheelchair users when performing a series of functional mobility tasks.

The vertical component of the castor forces during dynamic propulsion was gathered by instrumentation of the front castors of the wheelchair and through the push rim on the rear wheel of the wheelchair using a SmartWheel. The forces generated were then recorded and analysed to investigate the hypotheses stated above. Each subjectôs P:C Ratio was then calculated to establish whether the users reached or exceeded their capacity when performing such tasks and whether Rear Axle Position (RAP) was found to influence this.

3.3.2 Control Wheelchair

As there are no existing methods for testing dynamic wheelchair stability of a manual wheelchair, an instrument was developed. Consequently, a force sensitive castor was designed to look at this pattern in more detail and a wheelchair was dedicated to this purpose.

The chosen control wheelchair was a 17ò Quickie GPV, rigid frame lightweight wheelchair (Figure 3).This model was chosen as it was felt to be the most adaptable size for most test participants and could be configured to suit each individualôs needs. The wheelchair weighs 18kg with nearly 7kg of this weight due to attached testing equipment. The wheelchair is fitted with 25ò solid tyres to accommodate the SmartWheel technology and 5ò solid castors. There was a 3-degree camber on the rear wheels.

Figure 3 - The Quickie GPV Wheelchair

Each participant was set up in the control wheelchair and the wheelchair was set up to meet their needs, this included adjustment of the footplates and the backrest upholstery to support their posture. All other adjustments remained the same. Only one wheelchair could be used as the instrumentation was not interchangeable between wheelchairs.

3.3.3 Development of the Castor Force Transducer

Earlier work at Stanmore Clinical Research Facility (SCRF), using a wheelchair ergometer, showed that there was a substantial weight shift between castors and rear wheels during the propulsion cycle (Nicholson,G et al. 2006). The ergometer used force plates, as described by Wheatley et al (1980), beneath the front castors during propulsion on a roller system (Wheatley et al. 1980). The set up can be seen in Figure 4. There is no standardised method of measuring the forces directly through the castor it was therefore necessary to fit a force transducer to the castor stem in such a way that it measured the forces going through the castor. The design of the castor stem needed to be modified to allow for this which is outlined below. The calibration of this device was complex and is outlined in Appendix 2 to promote accurate repeatability of future studies.

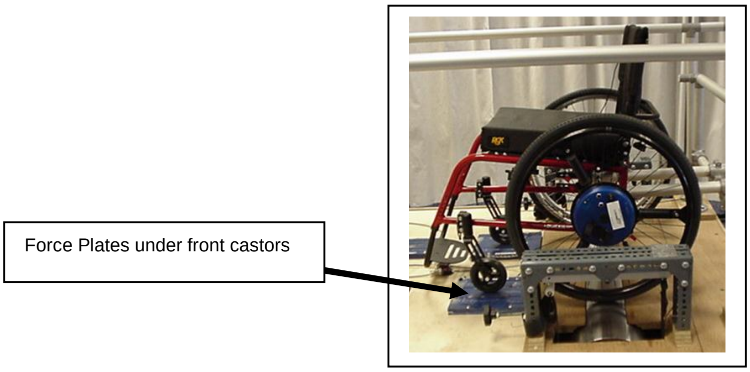


Figure 4 - The Wheelchair Ergometer

3.3.4 Instrumentation to measure Castor forces

In order to measure castor forces, force washers (Interface Inc, model LW2050-250 capacity 20Ibf) were placed on the stem of each castor between the castor fork and the castor bearing. Appendix 2 outlines the development of the castor construction.

The final castor construction (Figure 5) used for the experiment was composed of:

(a) **Bolt** - The bolt was used to establish the correct amount of tension on the spring during calibration.
(b) **Spring** - The spring allowed the castor stem assembly to be tightened.
(c) **Castor housing** - . The castor housing enclosed the linear bearing as described above to reduce the amount of axle friction inside the housing from the stem.
(d) **Rubber washer** - The rubber washer restricted the movement of the force transducer around the castor stem during testing and ensured that the cable was not damaged.
(e) **Force transducer** - A force transducer was purchased (Interface Inc, model LW2050-250 capacity 20Ibf) and attached to the PDA to collect data of weight distribution.
(f) **Original fork castor** - The original castor fork was used but not as free to swivel due to the limitations of the linear bearing.

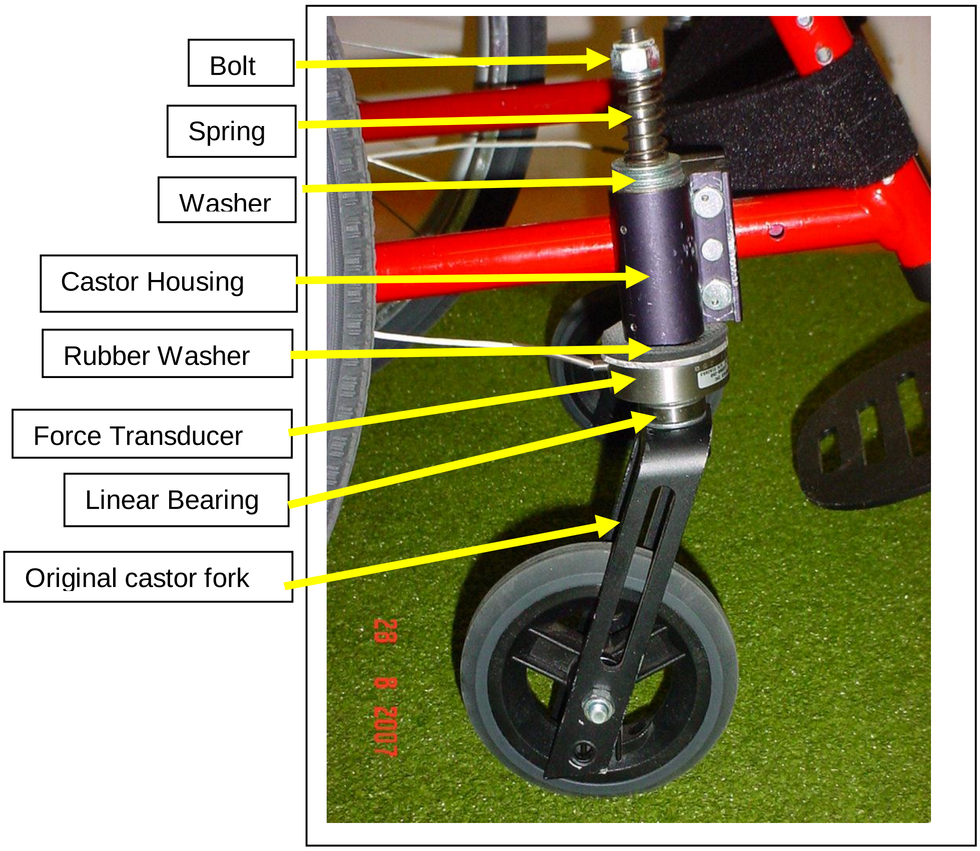


Figure 5 - Final castor configuration

3.3.5 Acquisition of Castor Data

The force washers were connected to a bridge amplifier (RS1210) that was constructed using an off-the-shelf printed circuit board (RS M12656). This circuit provided the means to control the excitation voltage to the strain gauge bridge of the force washer. It also allowed out of balance voltages to be zeroed. The two bridges (one for each castor) were powered by a voltage regulated battery power supply. The outputs were connected to a Dell Axim Pocket PC fitted with a National Instruments CF-6004 A-D converter. This proved to be a very simple and reliable method for collecting data from a mobile device, such as a wheelchair. The Pocket PC ran a Labview program developed for this project acquiring data at 250Hz per channel (Figure 6). See Appendix 3, for the detailed protocol on the Pocket PC.

Figure 6 - The pocket P.C running Lab View

The data was initially stored in binary form on the Pocket PC at a sampling rate of 250Hz. Once acquired, this was downloaded to a desktop computer where the binary files were converted into a readable format for data analysis. The complete data acquisition system was given the name ñTachyonò(Greek for ñSpeedyò) by the research team.

The additional data gathering equipment added 7kg of extra weight to the wheelchair. All calculations throughout the study were performed with this in place.

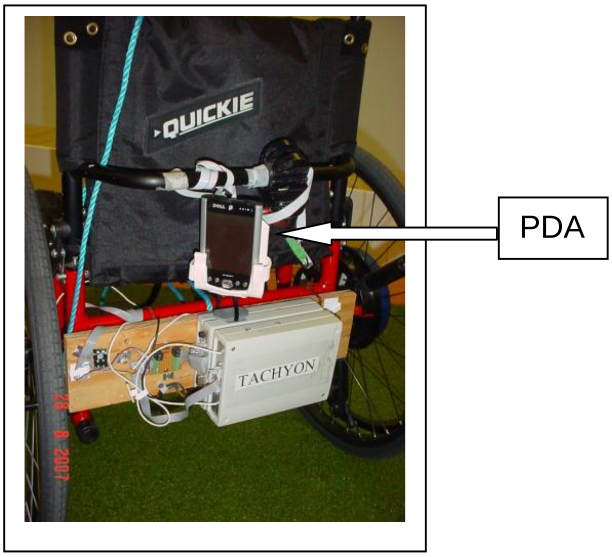
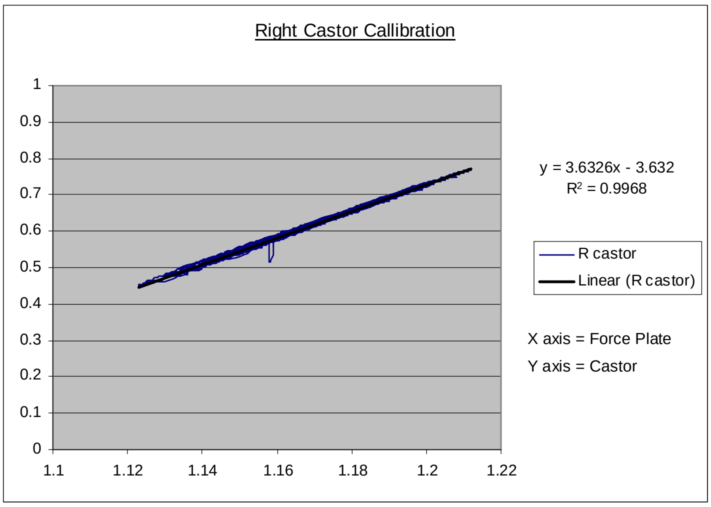


Figure 7 - The PDA on the wheelchair

3.3.6 Calibration of the castor force transducer

See Appendix 2 for full details on the calibration process. Figure 16 demonstrates the final calibration curve using the above construction. It can be seen that the return propulsion movement follows the same pattern as the forward movement demonstrating linearity and little hysteresis between the force plate signal and the force washer signal. There is a different calibration constant for the two force washers which, was attributable to the gain setting of the two bridge amplifiers.



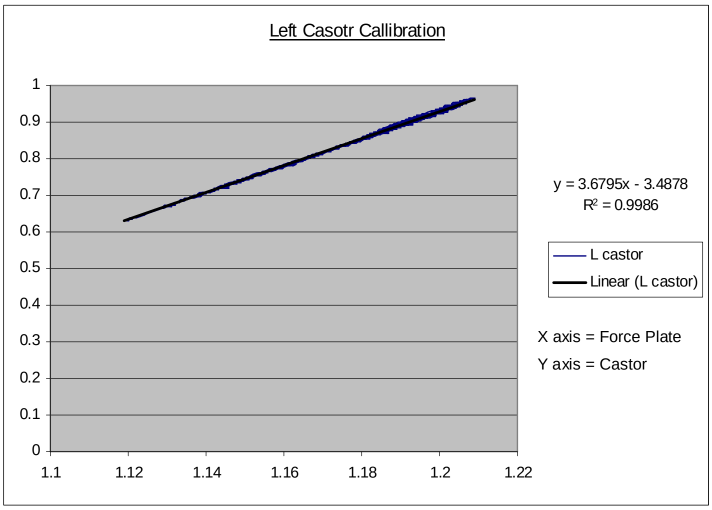


Figure 8 - Final castor calibration for right and left castor.

3.3.7 The SmartWheel for measurement of propulsion kinetics

The wheelchair was set up with a SmartWheel. The SmartWheel (Three Rivers Holdings) is an instrumented wheel, fitted to the wheelchair which gathers data on pushrim forces, moments, speed and acceleration. The SmartWheel is a calibrated and commercially available device and is shown in Figure 9.

Figure 9 - The SmartWheel

Vanlandewijck (2001) suggests that the SmartWheel can be mounted on the individualꞌs own wheelchair (Vanlandewijck et al. 2001). However, it was established that the SmartWheel required a certain axle receiver and therefore not all wheelchairs could receive the axle pin**.** This became apparent during the initial stages of the study. To overcome this barrier, the researcher liaised with SmartWheel manufacturers, who constructed a universal axle pin. As part of this a 25ꞌꞌ wheel with a solid tyre was also required for testing to match the construction of the SmartWheel. This was made by Three Rivers Holdings. This construction ensures that tyre pressure does not vary between subjects (Collinger et al. 2008).

The Smart Wheel was fitted with a 32Mb memory card that recorded the propulsion data. The data was then analysed using the 2005 version of the SmartWheel software (Three Rivers Holdings). The SmartWheel used in this study can measure forces in the range of ±155N and moments in the range of ±77Nm (R.A. Cooper *et al*. 1997). The forces are measured with a precision of 0.6 N and a resolution of 1N (R.A. Cooper *et al*. 1997). The moments are measured with a precision of 0.6 Nm and a resolution of 1Nm (R.A. Cooper *et al*. 1997). The wheel angle is measured from 0°-360°, with a precision of 0.18° and a resolution of 0.2° (R. A Cooper 1997).

A recent study by Cowan 2008 recommends that four parameters are specifically studied. These are taken from the ‘Guidelines on preservation of the Upper Limb’ (Cowan et al. 2008; Paralyzed Veterans of America 2005). These four parameters are velocity, force, push frequency and stroke length. The same key findings have been considered in this study in addition Peak Average Force was also analysed. One slight change is that Cowan analysed the peak resultant force. In this study the peak propulsion moment Mz was used[6]. The definitions of terms are outlined below as described by the SmartWheel Users Guide 2005 and WOWSUP 2006 (Nicholson,G et al. 2006) :

- **Stroke Angle –** This was defined as the angle travelled by the hand on the push rim from the point of contact to the point of release (in degrees). The average angle of the participant's push was recorded in degrees.
- **Cadence – or push frequency.** This is defined as how many times per second, on average, the participant pushes on the SmartWheel rim during an entire trial.
- **Velocity –** The average speed of the SmartWheel during each push. This can be used as an index of function. Average walking velocity is 1.4 m/s.
- **Peak Mz** – The peak propulsion moment that the participant applies to the SmartWheel during each push. This is the moment that turns the wheel (n/m).

6 This is essentially the same measure as the Peak Tangenital Force as Peak Mz is multiplied by a constant, which is the radius of the wheel i.e. 37.8 centimetres.

- **Peak Average Force Ratio –**The ratio between the peak force during a push, and the average force during a push. It provides an indication of how smoothly pushes are applied to the SmartWheelœs pushrim. A lower ratio indicates the peak force is more close to the average force, which can indicate a smoother push. Larger peak forces are associated with the development of upper extremity pain and dysfunction.

3.3.8 Terrain

The terrains included in this were study were those identified as part of the standard SmartWheel protocol for objective assessment (Three Rivers Holdings). These terrains are also referred to as functional mobility tasks throughout the study. These were developed by the SmartWheel international users group, which included members of ACDS as well as international key figures involved in manual wheelchair propulsion studies. The figure 8 test, outlined on the original protocol was not included, not only because it was seen as a skill assessment of manoeuvrability but the castor did not lend itself to testing in anything other than a straight line due to its difficulty with steer. However, an additional task of ascending a kerb was included, due to it being an everyday terrain encountered by the majority of active wheelchair users.

The functional mobility tasks included the following terrain types (
Table 2):

A. Straight push along 12m Lino on level ground
B. Straight push along 12m Astro Turf (Astro) on level ground (Springfield curl style from Lazy Lawn Artificial grasses)
C. Ascending a 1:12 Ramp ï this complies with building standards (maximum rise to run ï 1:12 or 5 º slope)
D. Ascending a 3ò Kerb (this involved braking once the kerb has been mounted)

It must be noted that start up parameters for a sloped surface represent the transition from level ground to a sloped surface (Cowan

2008) therefore in order to minimise this effect all subjects commenced the protocol with their castors already in position up the ramp.

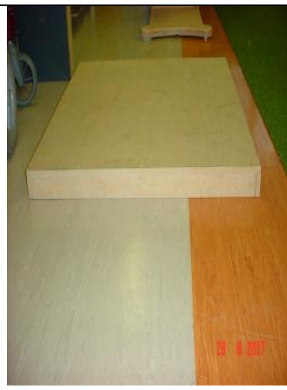
Terrain A & B	Terrain C	Terrain D
		

Table 2: Terrains

3.4 Subject Details

3.4.1 Recruitment Process

All subjects were experienced wheelchair users from Stanmore Spinal Cord Injuries Centre (SCIC), Stanmore UK.

3.4.2 Inclusion/Exclusion Criteria

Inclusion criteria for the participants were as follows:

- Subjects used a self propelling manual wheelchair was the primary mode of mobility;
- Subjects had been using a wheelchair for 2 or more years (this was classified as experienced);
- Have a spinal cord injury of the level T1 or below. This level was chosen as during initial trials the subjects with tetraplegia were unable to generate accurate readings from the SmartWheel.

Spinal Cord Injury level was determined in all participants using the American Spinal Injuries Association Classification (ASIA).

Participants were excluded from participating in the study if they reported a history of trauma to the upper limb or had experienced upper limb pain on pushing the wheelchair.

Seven men and one woman volunteered for the study. All participants met the inclusion criteria and provided written consent before they participated in the study (Appendix 1). See results section 4.1 for details of the participants.

3.5 Experimental Protocol

The full experimental procedure is outlined in Appendix 4. The Rear Axle Position (RAP) was adjusted for each subject, to give the most stable (back) and most tippy (forwards) position. The most stable position is defined as the most rear position that the axle can be on the axle plate. The most tippy position is defined as the most forwards position that the axle can be on the axle plate. The axle receiver is moved forwards and backwards within a plate mounted to the wheelchair, to adjust the stability of the wheelchair. Moving the RAP forwards will make the wheelchair more unstable/or tippy and moving it backwards will make the wheelchair more stable. The axle and its position are seen in Figure 10.

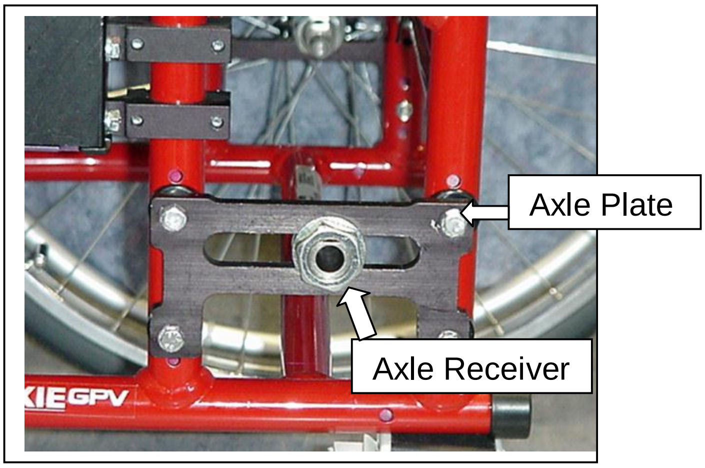


Figure 10 - The axle plate and receiver

In each axle position (most stable and most tippy) the following procedure was completed. This procedure also allowed for collection of data to calculate the Performance Capacity Ratio (2.10.1).

Stage One - Capacity

The MAX push (isometric push) - Each subject performed four isometric pushes by pushing the stationary wheelchair pushrim for 3 seconds with a rest period of 2 seconds in-between each MAX push. Wheels were prevented from rotation through the use of blocks at the front of the rear wheel, application of the brakes and by the wheelchair being positioned against a vertical surface.

This was performed with the users in the óPrimeôto push position as defined below, see 3.6.4. This data was used to calculate the capacity aspect of the P:C Ratio.

Stage Two - Performance

Participants performed a series of functional mobility tasks (3.3.8) at self selected speeds. Self selected velocity is thought to be important because of the way a person propels a wheelchair on an everyday basis maybe linked to shoulder pathology (Collinger et al. 2008). These were repeated three times with a rest period of 1 minute between each run. Although not measured, this rest period accounted for any muscular fatigue and diminished ability of the muscles to generate force over time (Rice et al. 2008). The course was completed in the same order for each of the participants. All participants completed the mobility course without difficulty, with the exception of one subject, who was unable to perform the ókerb runôin the tippy set up.

See Appendix 4 for detailed protocol.

3.6 Data Collection & Analysis

3.6.1 Synchronisation

The castor data was synchronised with the SmartWheel using the output from a tachometer, which was attached to the underside of the chair. This was activated as soon as the wheelchair moved, and this was taken as the start position for the castor data. As the SmartWheel was set to record on movement, this allowed synchronisation between the castor data and the SmartWheel data. In this instance the tachometer was not used to measure speed as the SmartWheel had the function to measure this.

3.6.2 Castor Data

Castor data was gathered using a bespoke programme written in LabVIEW 7.0, for a PDA which stored the data onto an SD card. Once all testing was completed, the raw binary files were downloaded onto a desktop computer and converted into a readable form, as seen in Figure 11. All readings were then converted from voltage units to force units (N) using the calibration data for the force washer/castor system and presented in an Excel spreadsheet format. A baseline was collected for each test to establish the output at zero loading and any offset was subtracted.

The calibration constants obtained from the data shown in Figure 8 were used along with the known calibration formulae for the force plates to convert the data from bridge amplifier output voltages into Newtons.

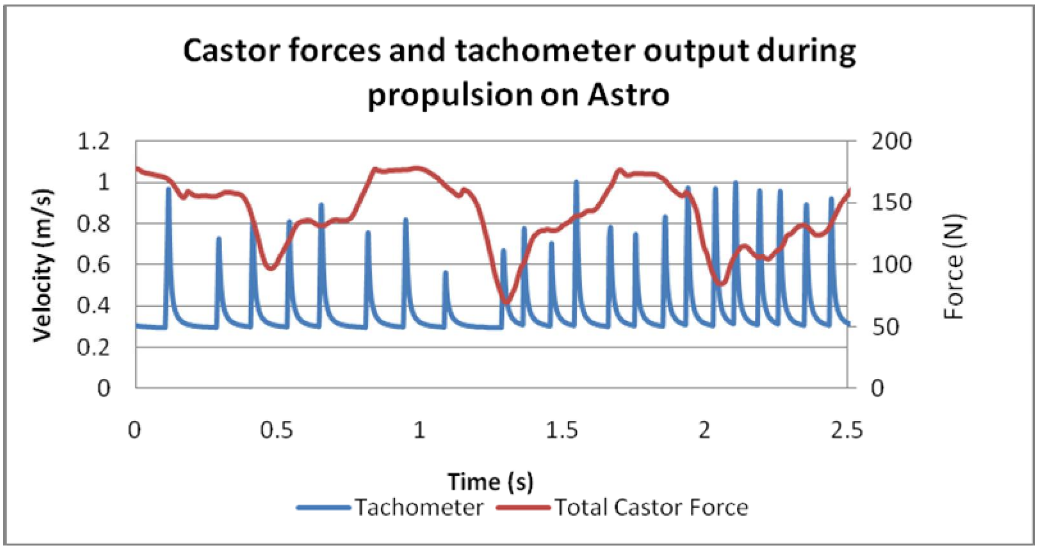


Figure 11 - Castor forces (kg) and tachometer output during propulsion on Astro

3.6.3 The SmartWheel Data

The data saved on the memory card from the SmartWheel was transferred to the desktop computer and analysed using the SmartWheel Analyser Software 2005 (Three Rivers Holding). The data was then transferred to a pre-prepared Microsoft template which had been written specifically for the programme (Nicholson 2005). This displayed a graph for analysis, as seen in Figure 12. The described biomechanical variables were then analysed (3.3.7).

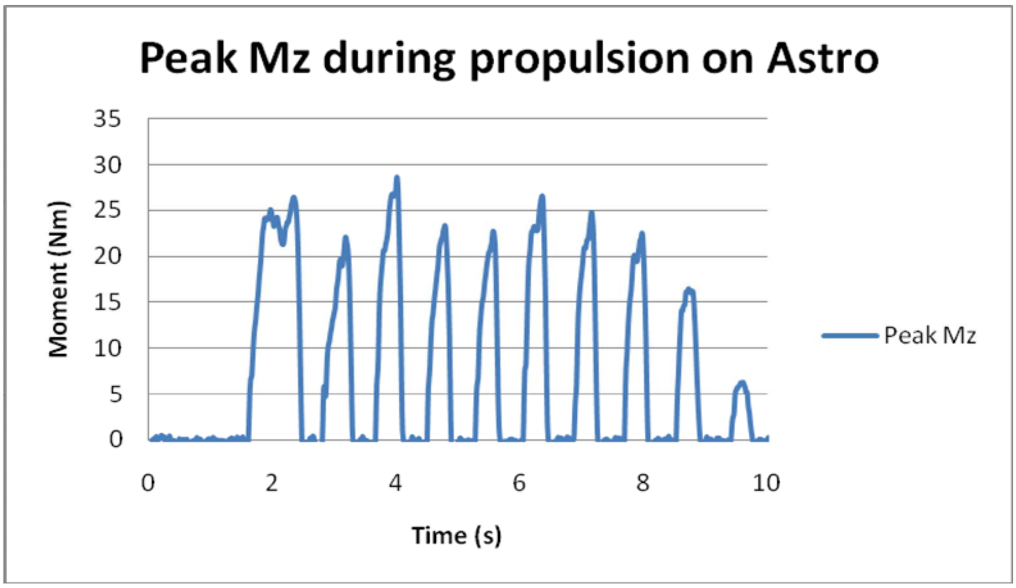


Figure 12 - Smart Wheel propulsion moment (Peak Mz) plotted against time (s)

The isometric data was used to represent the maximum capacity of the subjects whilst their data generated when propelling over the different terrains provided their performance data. Maximum values of Peak Mz were used to calculate this. The P:C ratio was then calculated using Equation 1.

$$\boldsymbol{P: C\ Ratio = \left(\frac{Performance}{Capacity}\right) \times 100\ \%}$$

Equation 1: Equation used to calculate the Performance: Capacity Ratio

It was decided to use a cut off point of 80% of a subjectô s capacity to indicate when a subject exceeds a safe level. When a subject reaches this level they were described as ó red-liningô (2.10.1).

3.6.4 SmartWheel and Castor Data Synchronisation

Once all data had been converted from the castor and the SmartWheel it was important for the propulsion cycle to be identified and defined. The same event had to be detected in both data recording systems in order for them to be synchronised for analysis. This included identifying:

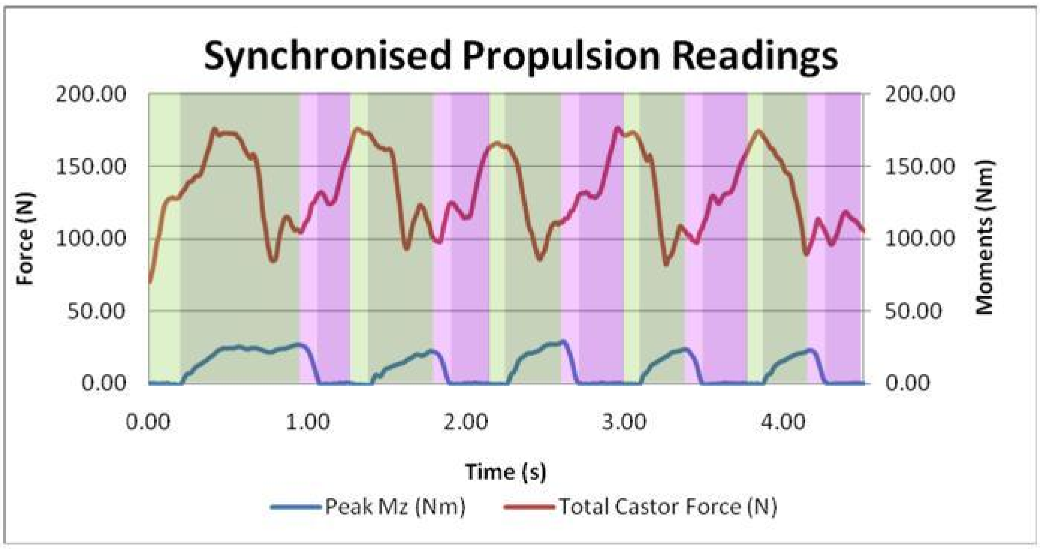
When wheel movement was first detected from the Tachometer, signalled by the PDA, and;

When the SmartWheel data indicated evidence of movement (velocity >0) using the 1/20s running average velocity parameter as defined in the SmartWheel Users Guide (2005).

By adopting this format and by consulting other studies (Vanlandewijck et al. 1994 & 2001; Kwarciak et al. 2009) the propulsion cycle was defined in this study as:

- **Primed to Push (Prime)** ï hands on the rim, push phase beginning, also described by Kwarciak et al as initial contact (Kwarciak et al. 2009).
- **Minimum Castor Force** ï lowest castor force recorded during the propulsion cycle.
- **Maximum Castor Force** ï highest castor force recorded during the propulsion cycle.
- **Push Phase** ï starts with a positive propulsion moment (Mz) and completed at hand release. Usually identified once contect is made with the push rim.
- **Recovery Phase** ï starts immediately after hand release and is completed at hand contact.
- **Hand Contact** ï when the hand makes contact with the rim
- **Hand Release** ï when the hand releases all contact with the rim

These principles were then adopted to analyse the data presented on the graphs. This is demonstrated in Figure 13 using the data generated when a subject performed on the Astro terrain with the RAP in the stable position.



Hand contact Push Phase Hand Release Recovery Phase

Figure 13 - The synchronised propulsion readings

3.6.5 Repeated measures

A test was conducted in order to identify if all data sets/terrain runs required analysis, or whether only one data set from each functional mobility task required selection. Researchers were unable to detect a difference between the three runs carried out in each functional mobility task. It could therefore be argued that there was no learning effect during the testing, but there was no possible way to identify any learning effect. Run number 2 was randomly selected as a result.

3.7 Statistics

Excel (spreadsheet software) and SPSS V13.0 (a statistical analysis software) programmes were used to analyse the data. The nature of the measurements were all continuous, therefore the data could be considered for parametric analysis. Each parameter was tested to establish if it was normally distributed using a Shapiro Wilks test. As all parameters were normally distributed , a Univariate Analysis Of Variance (ANOVA) was performed. This was completed to determine whether any significant difference in propulsion forces and castor loading occurred for extremes in chair tippiness, when performing the different functional mobility tasks. This test was used in order to effectively analyse the multiple data.

This was followed by a óPost Hoc Bonferroniô test to explore the interaction between the different terrains. A statistician was consulted to assist with identifying the most effective methods for reading and presenting the data. The significance level was set at ($p< 0.05$) for all statistical procedures.

3.8 Power Analysis

A power analysis was conducted to calculate the minimum sample size required to accept the outcome of a statistical test with a 90% level of confidence. A pilot dataset was collected to determine the variance in the measurements. The mean and the Standard Deviation was required from the sample size and this was taken for the Peak Mz data pushing on the lino, for one participant performing 5 repeated tests.

The sample size was determined using the first push Peak Mz for the mean tippy and the mean stable parameters and are represented by ε_0 and ε_1 respectively. The associated standard deviations for 5 trials were $\mathring{u}_0$ and $\mathring{u}_{1,}$ see Equation 2.

$$n = \frac{(\mathrm{u} + \mathrm{v})^2(\sigma_1^2 + \sigma_2^2)}{(\mu_1 - \mu_0)^2}$$

Equation 2: Equation to calculate the number of subjects needed taken from (Kirkwood & Sterne 2003)

Using Equation 2, the following values were used from a previous study: $\mathring{u}_{1\,=}$ std dev Mz first push on lino tippy = 2.8, $\mathring{u}_{0\,=}$ std dev Mz first push on lino stable = 2.8, u=1.29, v= 1.96 (u and v are values for normal distribution, for 90% power and a significance level of 0.5) , n = number of participants needed to detect an effect size of 6.9 with 90% power.

The value ε_1 ï ε_0 is the difference in the means i.e. the effect size and was set to be 30% of the mean of the stable data = 23. The effect size for clinical significance is therefore > 6.9. Substituting in the values yields:

$$n = \frac{(1.29 + 1.96)^2(2.8^2 + 2.8)^2}{6.9^2} = 3.5$$

Equation 3: Equation to calculate the number of subjects needed for this study using inputs from WOWSUP

It was decided that a sample of 8 participants would be recruited to represent the range of different functional capacities of people with SCI.

4 Results

The results of the research will be presented separately; with reference made to the three hypotheses used (3.1.1, 3.1.2,3.1.3). Each hypothesis is analysed for the parameters *primed to push*, *min push* and *max push* with references to the *first push*, the *acceleration phase*, (also known as *start up phase*) and the *steady state phase*. The P:C Ratio is also calculated for each person, under each test scenario and the results are presented. The results will be discussed in Chapter 5 of the thesis.

4.1 Subjects

Seven men and one woman volunteered for the study. All participants met the inclusion criteria. Table 3 provides the characteristics of the subjects.

User	Injury	ASIA	Time since Injury (Years)	Gender	Age (Years)	Weight (Kg)
1	T12	C	35	Male	54	70.0
2	T5	A	6	Male	52	71.0
3	T11	A	10	Male	45	73.3
4	L3	A	3	Female	43	77.4
5	T6	A	12	Male	41	72.0
6	T12	A	2	Male	27	83.0
7	T8	A	11	Male	37	64.6
8	T9	A	12	Male	49	72.0

Table 3 - Characteristics of subjects included in the study

4.2 Hypothesis 1: Castor Forces and the influence of RAP

H0: There is no significant difference in the castor forces generated during straight line functional mobility tasks when the rear axle is moved forward (tippy) compared to the most stable (rearmost) position.

H1: There are significantly lower castor forces generated during straight line functional mobility tasks when the rear axle is moved forward (tippy) compared to the most stable (rearmost) position.

Prime push

The average castor forces generated in the prime position (for the purposes of this thesis this is considered to be equivalent to static sitting with hands on the pushrim) are , as expected, lower when the RAP is set forwards in the tippy position ($p < .001$) (see Table 4). This is where the hands are on the rim about to commence the propulsion cycle.

Min castor forces

The average minimum castor forces are significantly lower in the tippy position for the first push (<0.001), second push (<0.001) and the first push of the steady state (p Ò0.004) as detailed in Table 4. However, there is no significant difference throughout the remainder of the propulsion cycle.

Max castor forces

Table 4 shows that max castor forces are significantly greater in the stable position compared to the tippy position ($p < 0.001$).

Terrain Stability	Lino Stable	Lino Tippy	Astro Stable	Astro Tippy	Ramp Stable	Ramp Tippy	RAP F(1,32)	RAP p Value
Prime 1	134.6	81.0	169.8	108.3	139.3	74.6	63.2	<.001
Min 1	61.8	28.5	68.1	39.9	56.3	25.6	15.0	<.001
Max 1	234.3	172.8	284.3	230.0	260.8	153.5	35.7	<.001
Prime 2	145.2	87.5	160.2	121.4	141.0	75.0	50.06	<.001
Min 2	73.1	33.2	68.2	46.5	61.4	17.3	16.48	<.001
Max 2	226.5	198.7	283.9	247.8	279.0	201.4	17.33	<.001
Prime 3	137.1	96.6	161.8	102.7	142.2	84.3	41.29	<.001
Min 3	71.0	40.9	60.6	49.1	49.1	31.9	3.92	NS
Max 3	240.1	203.5	297.7	262.0	277.6	230.2	12.08	<0.001
Prime L1	134.9	93.0	171.4	130.7	147.7	73.2	23.99	<0.001
Min L1	73.3	39.6	65.9	44.2	56.4	19.5	9.55	0.004
Max L1	238.0	194.2	302.8	241.9	274.2	191.7	22.41	<.001
Prime L2	143.9	108.9	178.5	124.9	132.7	88.9	13.28	0.001
Min L2	80.0	55.5	63.1	49.7	58.8	24.3	4.13	NS
Max L2	231.8	174.4	291.4	241.4	270.4	223.2	121.67	<.001
Prime L3	140.9	85.5	164.2	125.1	155.2	108.2	15.15	<.001
Min L3	98.5	50.2	68.9	57.6	87.2	43.7	6.50	NS
MaxL3	227.8	171.4	275.4	238.8	259.4	206.0	19.43	<.001

Table 4 – Average Total Castor Forces (n) for Prime, min and max push over all terrains and their significance in relation to RAP (NS = Not significant) yellow highlighting shows significant difference in RAP in relation to each push. Green shading refers to first 3 pushes and blue shading refers to last 3 pushes.

These results show that there is a significant difference between the castor forces when the RAP is adjusted ($p < 0.05$). The null hypothesis can be rejected as moving the rear axle forward (less stable) reduces the amount of castor forces during functional mobility tasks. This is highlighted in Figure 14, which consistently shows the stable RAP (red) being greater than the tippy RAP (blue). It can be concluded that moving the rear axle forward reduces the amount of castor force during functional mobility.

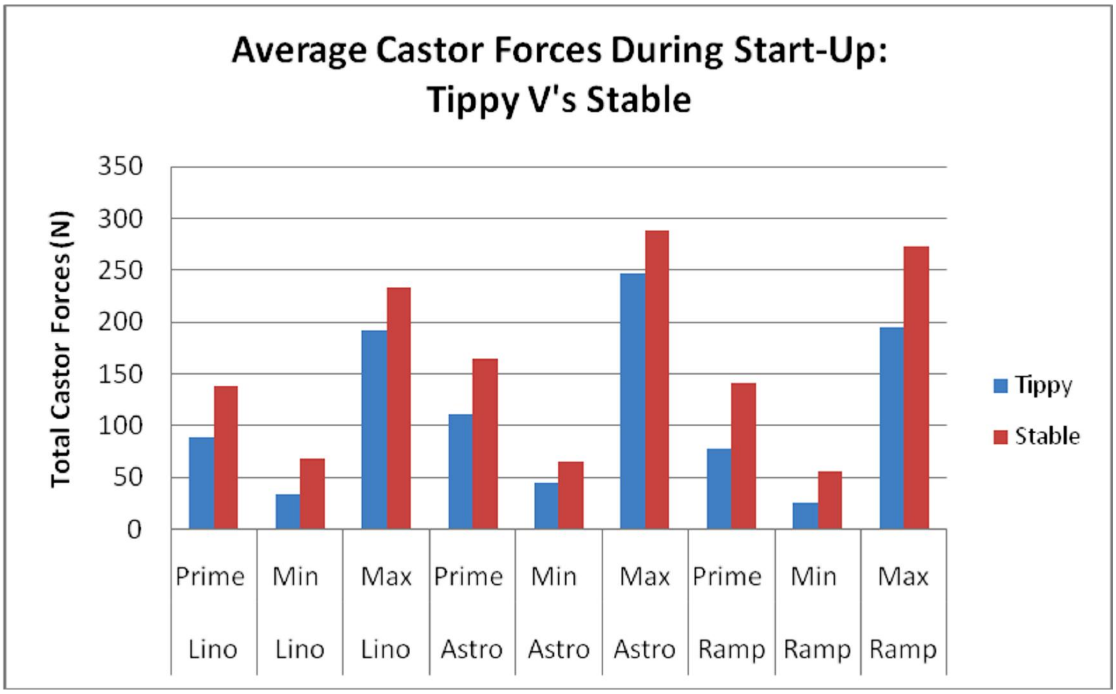


Figure 14: Graph showing the comparison between average castor forces generated during the start-up phase (N) when the RAP is set in the most stable and the most tippy position.

4.3 Hypothesis 2: Castor Forces and the influence of terrain

H0: There is no significant difference in the castor forces generated when the rear axle is moved forwards (tippy) during straight line functional mobility over different terrains.

H1: The castor forces are lower when the rear axle is moved forward (tippy) during straight line functional mobility over different terrains.

Prime push

Prime push shows significant differences between the Astro surface and the ramp for the first push (p=0.005), the second push (p=0.004) and also the first push of the steady state (p=0.014) with Astro being higher than the ramp. However, it was not consistently significant. This pattern is replicated for the lino compared with Astro, again with

Astro generating higher castor forces (Table 5). When lino is compared to the ramp there is no significant difference in the castor force readings.

Min castor forces

Min castor forces generate no significant difference for any of the terrains.

Max castor forces

Significant differences in max castor force (p Ò0.009) can be seen between the terrains. However, when the individual comparisons are examined it is clear that there is only a significant difference between lino and Astro, with Astro generating the higher castor forces. There is one instance of a significant difference between the lino and ramp terrains, this is seen half way through the steady state ($p = 0.005$). The lack of a significant difference between the ramp and lino conditions could be due to the large variation in castor force max. When the ócastor force maxô measurement is taken, there is no propulsion force at this point of the propulsion cycle as the hand is removed from the rim at the end of the stroke. It is at this point in the cycle that the trunk is at maximum flexion; which is the reason for the maximum castor force reading.

Terrain Stability	Lino Stable	Lino Tippy	Astro Stable	Astro Tippy	Ramp Stable	Ramp Tippy	Terrain F Value	Terrain p Value	Terrain Lino Astro	Terrain Lino Ramp	Terrain Astro Ramp
Prime1A	134.6	81.0	169.8	108.3	139.3	74.6	$F(2,32) = 7.6$	0.002	0.003	NS	0.005
Min1A	61.8	28.5	68.1	39.9	56.3	25.6	$F(2,32) = .111$	NS	NS	NS	NS
Max1A	234.3	172.8	284.3	230.0	260.8	153.5	$F(2,32) = 6.17$	0.005	0.003	NS	0.013
Prime2A	145.2	87.5	160.2	121.4	141.0	75.0	$F(2,32) = 5.83$	0.007	0.022	NS	0.004
Min2A	73.1	33.2	68.2	46.5	61.4	17.3	$F(2,32) = .76$	NS	NS	NS	NS
Max2A	226.5	198.7	283.9	247.8	279.0	201.4	$F(2,32) = 5.55$	0.009	0.001	NS	NS
Prime3A	137.1	96.6	161.8	102.7	142.2	84.3	$F(2,32) = 1.63$	NS	NS	NS	NS
Min3A	71.0	40.9	60.6	49.1	49.1	31.9	$F(2,32) = .287$	NS	NS	NS	NS
Max3A	240.1	203.5	297.7	262.0	277.6	230.2	$F(2,32) = 7.48$	0.002	<.001	NS	NS
PrimeL1A	134.9	93.0	171.4	130.7	147.7	73.2	$F(2,32) = 6.35$	0.005	0.015	NS	0.014
MinL1A	73.3	39.6	65.9	44.2	56.4	19.5	$F(2,32) = .773$	NS	NS	NS	NS
MaxL1A	238.0	194.2	302.8	241.9	274.2	191.7	$F(2,32) = 6.52$	0.004	0.004	NS	NS
PrimeL2A	143.9	108.9	178.5	124.9	132.7	88.9	$F(2,31) = 2.72$	NS	NS	NS	NS
MinL2A	80.0	55.5	63.1	49.7	58.8	24.3	$F(2,32) = .79$	NS	NS	NS	NS
MaxL2A	231.8	174.4	291.4	241.4	270.4	223.2	$F(2,32) = 9.20$	0.001	<.001	0.005	NS
PrimeL3A	140.9	85.5	164.2	125.1	155.2	108.2	$F(2,31) = 2.36$	NS	NS	NS	NS
MinL3A	98.5	50.2	68.9	57.6	87.2	43.7	$F(2,32) = .334$	NS	NS	NS	NS
MaxL3A	227.8	171.4	275.4	238.8	259.4	206.0	$F(2,32) = 6.52$	0.004	<.001	NS	NS

Table 5 - Castor Forces for Prime, minimum and max push over all terrains and their significance in relation to terrain (NS= Not significant and green highlighting shows significant difference in terrain)

When looking at the castor forces across the different terrains there is a less straightforward series of results than with the RAP (Table 5). There were minimal differences seen for the castor forces when subjects performed on the ramp, compared to lino. The main differences were found between lino and Astro which are predominantly associated with the prime and the max push parameters.

4.4 Hypothesis 3: SmartWheel Data

H0: There are no significant differences in individual SmartWheel parameters (Peak Mz, velocity, stroke angle and cadence) when the rear axle is moved forwards during straight line functional mobility over different terrain.

H1: There are significant differences in individual SmartWheel parameters (Peak Mz, velocity, stroke angle and cadence) when the rear axle is moved forwards during straight line functional mobility over different terrain.

Terrain Stability	Lino Stable	Lino Tippy	Astro Stable	Astro Tippy	Ramp Stable	Ramp Tippy	Terrain F Value	Terrain p Value	Terrain Lino Astro	Terrain Lino Ramp	Terrain Astro Ramp	RAP F Value	RAP p Value
StrAng1	79.5	80.9	79.5	70.2	86.8	85.8	$F(2,32) = 0.983$	NS	NS	NS	NS	$F(1,32) = .414$	NS
StrAng2	81.2	89.8	75.2	87.8	91.1	88.6	$F(2,32) = 0.300$	NS	NS	NS	NS	$F(1,32) = 1.38$	NS
StrAng3	80.4	86.8	83.5	88.9	92.7	82.5	$F(2,32) = .136$	NS	NS	NS	NS	$F(1,32) = 0.02$	NS
StrAng12	80.4	85.4	77.3	79.0	88.9	87.2	$F(2,32) = .85$	NS	NS	NS	NS	$F(1,32) = 0.33$	NS
StrAng13	80.4	85.8	79.4	82.3	90.2	85.7	$F(2,32) = .576$	NS	NS	NS	NS	$F(1,32) = 0.212$	NS
StrAngsteady	74.2	73.8	73.1	109.2	92.3	76.2	$F(2,32) = 2.90$	NS	NS	NS	NS	$F(1,32) = 1.07$	NS
Cad1	0.4	0.4	0.4	0.4	0.5	0.4	$F(2,32) = 0.98$	NS	NS	NS	NS	$F(1,32) = 0.41$	NS
Cad2	0.4	0.5	0.4	0.5	0.5	0.5	$F(2,32) = 0.300$	NS	NS	NS	NS	$F(1,32) = 1.38$	NS
Cad3	0.4	0.5	0.4	0.5	0.5	0.4	$F(2,32) = 0.14$	NS	NS	NS	NS	$F(1,32) = 0.02$	NS
Cad12	0.4	0.4	0.4	0.4	0.5	0.5	$F(2,32) = 0.85$	NS	NS	NS	NS	$F(1,32) = 0.33$	NS
Cad13	0.4	0.4	0.4	0.4	0.5	0.4	$F(2,32) = 0.58$	NS	NS	NS	NS	$F(1,32) = 0.21$	NS
Cadsteady	0.4	0.4	0.4	0.6	0.5	0.4	$F(2,32) = 2.90$	NS	NS	NS	NS	$F(1,32) = 1.07$	NS
Vel1	0.8	0.6	0.6	0.5	0.7	0.6	$F(2,32) = 0.77$	NS	NS	NS	NS	$F(1,32) = 8.21$	0.007
Vel2	1.3	1.2	1.0	1.0	1.0	0.9	$F(2,32) = 14.57$	<.001	<.001	<.001	NS	$F(1,32) = 0.73$	NS
Vel3	1.5	1.5	1.2	1.1	1.1	1.0	$F(2,32) = 42.17$	<.001	<.001	<.001	0.049	$F(1,32) = 1.90$	NS
Vel12	1.0	0.9	0.8	0.7	0.8	0.7	$F(2,32) = 4.21$	0.024	0.004	0.008	NS	$F(1,32) = 4.66$	0.038
Vel13	1.1	1.0	0.9	0.9	0.9	0.8	$F(2,32) = 12.04$	<.001	<.001	<.001	NS	$F(1,32) = 5.01$	0.032
Velsteady	1.8	2.0	1.3	1.0	1.1	1.0	$F(2,32) = 49.31$	<.001	<.001	<.001	NS	$F(1,32) = 0.72$	NS
PeakMz1	22.1	24.5	27.5	23.9	28.4	32.4	$F(2,32) = 1.46$	NS	NS	0.049	NS	$F(1,32) = 0.43$	NS
PeakMz2	20.5	23.9	27.0	26.8	34.6	35.1	$F(2,32) = 3.40$	0.046	NS	0.003	NS	$F(1,32) = 0.38$	NS
PeakMz3	18.0	21.9	25.7	26.9	28.9	28.3	$F(2,32) = 1.43$	NS	NS	NS	NS	$F(1,32) = 0.23$	NS
PeakMz12	21.3	24.2	27.2	25.3	31.5	33.8	$F(2,32) = 2.82$	0.074	NS	0.006	NS	$F(1,32) = 0.466$	NS
PeakMz13	20.2	23.4	26.7	25.8	30.6	31.9	$F(2,32) = 2.69$	0.084	NS	0.007	NS	$F(1,32) = 0.453$	NS
PeakMzsteady	11.3	13.5	19.5	18.1	25.3	27.8	$F(2,32) = 11.14$	<.001	0.076	<.001	0.025	$F(1,32) = 0.343$	NS
Peakaveforce1	1.6	1.8	1.6	1.8	1.7	2.1	$F(2,32) = 0.926$	NS	NS	NS	NS	$F(1,32) = 7.05$	0.012
Peakaveforce2	1.5	1.5	1.5	1.6	1.7	1.7	$F(2,32) = 1.20$	NS	NS	NS	NS	$F(1,32) = 0.332$	NS
Peakaveforce3	1.4	1.5	1.6	1.6	1.6	1.7	$F(2,32) = 1.422$	NS	NS	0.018	NS	$F(1,32) = 7.49$	0.01
Peakaveforce12	1.6	1.7	1.6	1.7	1.7	1.9	$F(2,32) = 1.189$	NS	NS	NS	NS	$F(1,32) = 4.26$	0.047
Peakaveforce13	1.5	1.6	1.6	1.6	1.7	1.9	$F(2,32) = 1.368$	NS	NS	0.044	NS	$F(1,32) = 5.68$	0.023
Peakaveforcesteady	1.5	1.6	1.5	1.4	1.5	1.6	$F(2,32) = 3.80$	0.033	NS	NS	NS	$F(1,32) = 3.22$	NS

Table 6 - Table showing analysed SmartWheel variables over all terrains and their significance in relation to RAP and Terrain (NS= Not significant. Green highlighting shows significant difference in RAP, yellow highlighting shows significant difference in terrain)

4.4.1 Stroke Angle and Cadence

From the data gathered Stroke Angle and Cadence are not affected by terrain or stability. Data can be seen in Table 6.

4.4.2 Velocity

Velocity was higher when subjects performed on lino compared to the other terrains, this was apparent in both the stable and tippy set-up ($p<0.001$), see Figure 15**.** This is also supported by Table 8 which shows the average velocity in metres per second over the different terrains.

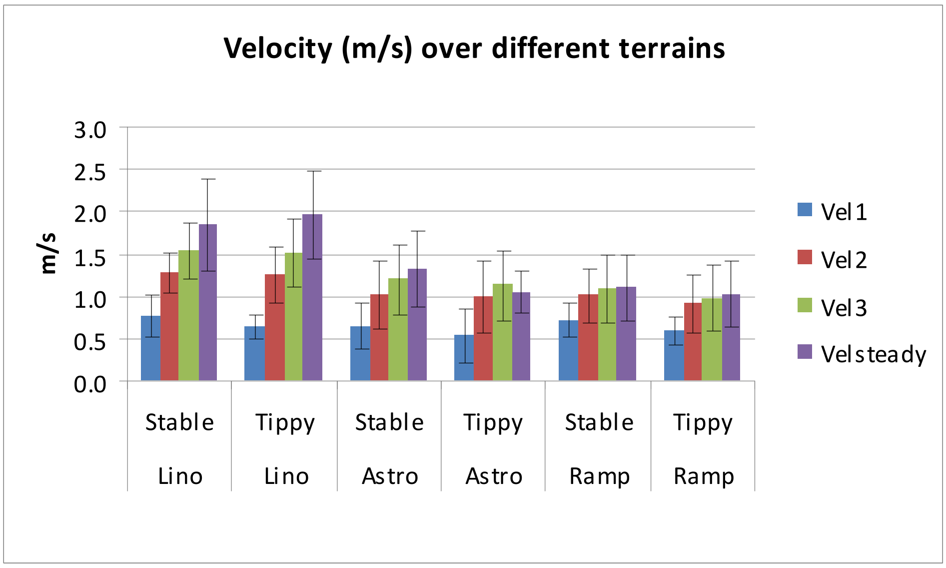


Figure 15 - Graph showing the subjects velocities and the comparison of mean (s.d.) velocities for the 3 functional mobility tasks for the two wheelchair stability configurations (stable, tippy).

There is variability in the data and this is consistent for all the terrains reflecting both individual abilities and the small number of subjects included in this study. There is more variability seen in the steady state part of the push when performing on the lino and this may reflect individual ability and technique.

For all conditions following the first push, velocity is influenced by terrain for lino compared to Astro, and also lino compared to the ramp ($p<0.001$), see Table 7 for individual p values. Aside from the third push ($p = 0.049$) there is no significant difference found between the Astro and ramp terrains.

	p Value	Lino Astro	Lino Ramp	Astro Ramp
Velocity 1	NS	NS	NS	NS
Velocity 2	<.001	<.001	<.001	NS
Velocity 3	<.001	<.001	<.001	0.049
Vel steady state	<.001	<.001	<.001	NS

Table 7 - Table showingthe significance of terrains on velocity during the first three pushes and steady state (NS=Not significant)

In relation to the RAP, Velocity was one of the only SmartWheel parameters to be influenced by stability, showing an increase in velocity when subjects performed in the more stable position (see Table 8). However, this was only in half of the scenarios with the most significant being at first push ($p = .007$) (see Table 6). And the variability between the different set ups is not as apparent.

Terrain	Lino	Lino	Astro	Astro	Ramp	Ramp
Stability	Stable	Tippy	Stable	Tippy	Stable	Tippy
Vel1	0.8	0.6	0.6	0.5	0.7	0.6
Vel2	1.3	1.2	1.0	1.0	1.0	0.9
Vel3	1.5	1.5	1.2	1.1	1.1	1.0
Velsteady	1.8	2.0	1.3	1.0	1.1	1.0

Table 8 - Table to show the effect of RAP on the average velocity (m/s) over all terrains

4.4.3 Peak Mz

This study shows that the position of the rear axle does not significantly affect the propulsion forces of the wheelchair on any of the terrains. However, although RAP showed no significant difference, terrain did, demonstrating a significant difference between the forces necessary to propel over these different terrains.

Peak Mz is significantly affected by terrains in the steady state phase of the push ($p < 0.01$), with the ramp requiring a higher peak moment than the Astro, which in turn requires a higher moment than the lino. This is highlighted in Figure 16 and Table 9, which shows all pushes, aside from the third push, are significant between the lino and the ramp.

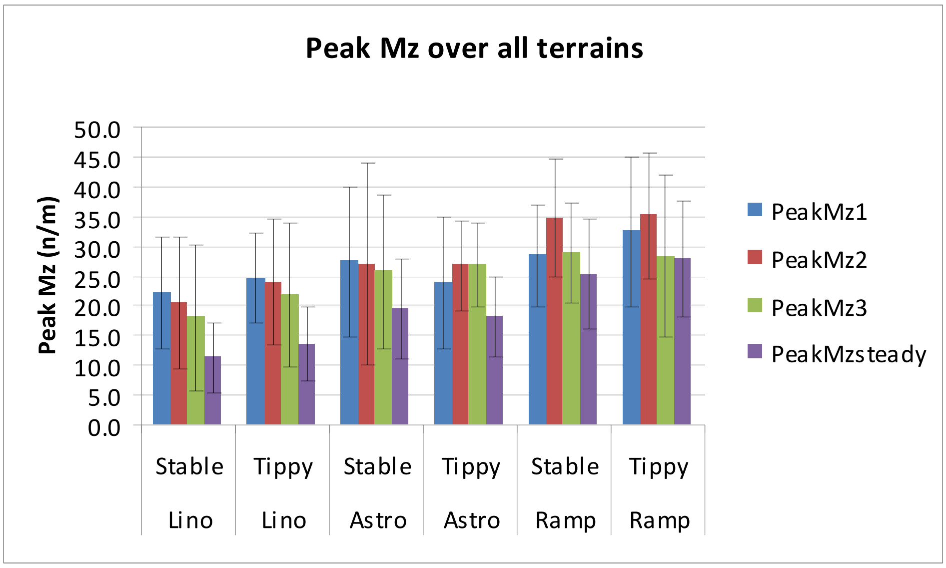


Figure 16 - Graph showing the subjects Peak Mz and the comparison of mean (s.d.) Peak Mz for the 3 functional mobility tasks for the two wheelchair stability configurations (stable, tippy).

Variability in the data can be seen over all terrains reflecting individual performance. What can be seen is that the steady state part of the push is less variability possibly due to a leveling off of individual performance. Again on the ramp, the variability is consistent between the wheelchair set up and the pushes supporting the results of the ramp and the need for the same level of push throughout the task.

	p Value	Lino Astro	Lino Ramp	Astro Ramp
Peak Mz 1	NS	NS	0.049	NS
Peak Mz 2	0.046	NS	0.003	NS
Peak Mz 3	NS	NS	NS	NS
Peak Mz steady	<.001	NS	<.001	0.025

Table 9 - Table showing the significance of terrain on Peak Mz during the first three pushes and Steady State (NS= Not significant).

4.4.4 Peak Average Force Ratio

Peak Average Force shows no significant difference over the different terrains but is one of the only parameters to be influenced by stability (< 0.047) see Table 6. There is also less variability seen in the data (Figure 17).

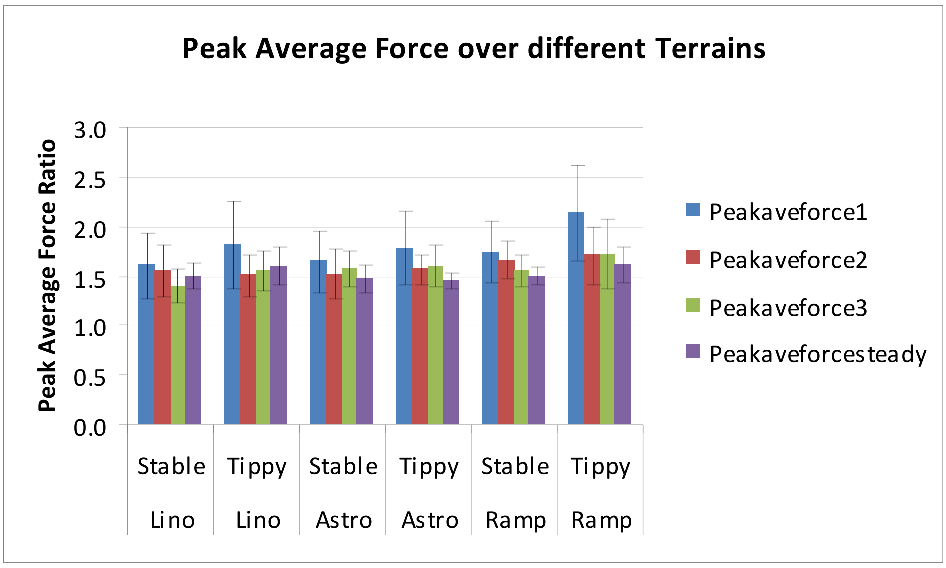


Figure 17 - Graph showing the subjects Peak Average Force and the comparison of mean (s.d.) Peak Average Force for the 3 functional mobility tasks for the two wheelchair stability configurations (stable, tippy).

Table 10 shows a greater reading of Peak Average Force in the tippy position compared to the stable position indicating that subjects performed smoother pushes when performing terrains in the more stable position

	RAP p Value
Peak aveforce 1	0.012
Peak aveforce 2	NS
Peak aveforce 3	0.01
Peak aveforce steady	NS

Table 10: Table showing the effect of RAP on Peak Average Force during the first three pushes and Steady State (NS= Not significant).

The results show that certain SmartWheel parameters are affected by RAP. Each Smart Wheel parameter will be presented individually along with the results found.

4.5 Ascending a Kerb

When performing the Kerb test, data were collected from the SmartWheel. The castor data was not considered to have any relevance in this test as the castors were off the ground at the crucial part of the measurements. The propulsion parameters for the wheelchair user performing the kerb test can be split into three push phases, which is also demonstrated in Figure 18**.**

- **Push 1** ï initial moment generated to flip the castors up the kerb.
- **Push 2** ï moment generated to get over the kerb
- **Push 3** ï moment required to stop the wheelchair moving forwards (braking). This would not be usual when ascending a kerb functionally as most wheelchair users would continue along the pavement.

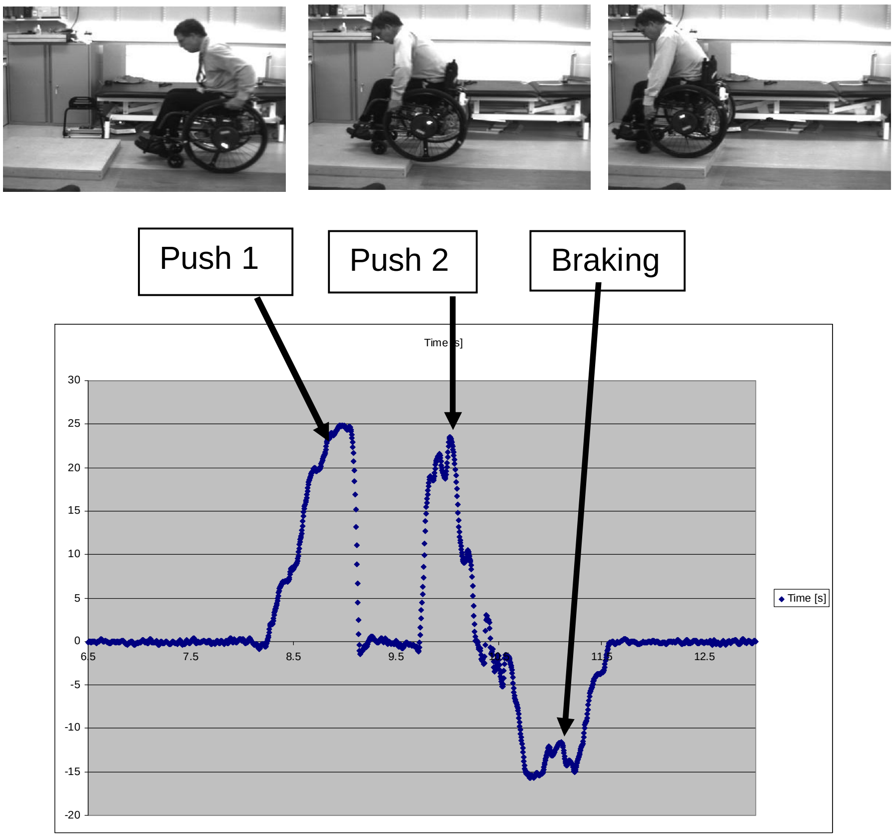


Figure 18 - SmartWheel Propulsion Moment Mz for the kerb plotted against time

Peak Mz shows a significant difference (p =.023) in Push 1 between the tippy and stable set up with a greater Peak Mz performed to flip the castors with the wheelchair in the stable position. Push 2 analysis found no significant difference in Peak Mz between the two RAP's.

4.6 P:C RATIO

To exceed a P:C Ratio or ‘red line’, users have to reach and exceed 80% of their capacity. This is represented by colour coding the results – see key below.

4.6.1 P:C Ratio for Lino

When performing on the lino two subjects exceeded their P:C ratio, see Table 9.

Subject	Peak Mz S [N]	(N)	P:C Lino S 1st	P:C Lino T 1st	PC Lino S SS	PC lnio T SS
1	47	32	48%	67%	16%	23%
2	64	43	27%	40%	11%	19%
3	69	50	10%	64%	7%	33%
4	38	40	35%	51%	38%	36%
5	40	57	86%	50%	40%	34%
6	68	50	44%	49%	16%	21%
7	31	56	100%	75%	71%	43%
8	64	49	29%	32%	12%	15%

Key: S – Stable, **T** – Tippy, **1st** – Start up, **SS** – Steady State

Up to 49% of capacity (green)

Up to 79% of capacity (amber)

Up to 80% and above capacity (red lining)

Table 11- Table showing the calculated P:C Ratio over lino terrain in the stable and tippy RAP

Subjects were more likely to perform at less than 50% of their capacity when the RAP was in the most stable position; whereas in the ‘tippy’ set-up subjects were more likely to exceed 50%. However, two subjects did exceed their P:C ratio in the stable set-up and ‘red lined’ (subjects 5 & 7). These subjects both have a higher level of spinal cord injury (Table 3).

4.6.2 PC Ratio – Astro

There was an increase in the number of subjects who exceeded their P:C Ratio whilst performing the run on the Astro terrain compared to the lino, see Table 12.

Subject	Peak Mz S [N]	Peak Mz T (N)	P:C Astro S1st	P:C Astro Tst	PC Astro S SS	PC Astro T SS
1	47	32	57%	81%	32%	31%
2	64	43	36%	45%	34%	41%
3	69	50	14%	68%	5%	57%
4	38	40	80%	68%	73%	51%
5	40	57	113%	69%	79%	17%
6	68	50	49%	76%	26%	44%
7	31	56	172%	100%	62%	64%
8	64	49	35%	46%	29%	38%

Key: S ï Stable, **T** ï Tippy, **1st** ï Start up, **SS** ï Steady State

Up to 49% of capacity (green)

Up to 79% of capacity (amber)

Up to 80% and above capacity (red lining)

Table 12 - Table showing the calculated P:C Ratio over Astro terrain in the stable and tippy RAP

Three subjects (4,5 & 7) exceeded their P:C Ratio with the RAP positioned in the most stable position and two subjects exceeded in the more tippy set up (1&7). Subject 7 exceeded in both the tippy and the stable set up. All óred liningô occurred in the Start up phase of the run (first three pushes) and again was more likely to occur in the more stable set up.

4.6.3 PC Ratio – Ramp

Table 13 shows that the ramp present considerable difficulties compared with the other surfaces. However, it can be seen that there is a shift away from subjects exceeding their P:C Ratio with the RAP in the most stable position and it is seen more frequently when the RAP is in the tippy position.

When subjects ascend the ramp, there is a shift from red liningôat start-up phase to it occurring during the steady state part of the run. There is also a higher incidence of subjects performing within the amber range (up to 79% of their capacity) on the ramp. This suggests that subjects need to perform at an elevated level throughout the task in order to prevent rolling backwards down the ramp, making this a more difficult terrain for subjects to perform.

Subject	Peak Mz S [N]	Peak Mz T (N)	P:C Ramp S1st	P:C Ramp Tst	PC Ramp S SS	PC Ramp T SS
1	47	32	76%	121%	48%	108%
2	64	43	50%	59%	43%	55%
3	69	50	39%	59%	31%	34%
4	38	40	79%	58%	69%	51%
5	40	57	93%	70%	87%	50%
6	68	50	74%	90%	60%	79%
7	31	56	60%	97%	44%	74%
8	64	49	63%	64%	24%	37%

Key: S ï Stable, **T** ï Tippy, **1st** ï Start up, **SS** ï Steady State

Up to 49% of capacity (green)

Up to 79% of capacity (amber)

Up to 80% and above capacity (red lining)

Table 13 - Table showing the calculated P:C Ratio over the Ramp terrain in the stable and tippy RAP

4.6.4 P:C Ratio - Kerb

When performing on the kerb subject 7 exceeded their P:C Ratio in both the tippy and stable set-ups. Subjects 5 & 7 again red lined in the stable position, along with subject 2, who was unable to complete the kerb task with the RAP in the Tippy position (Table 14).

Subject	Peak Mz S [N]	Peak Mz T (N)	P:C Kerb S1st	P:C Kerb Tst
1	47	32	50%	85%
2	64	43	87%	NA
3	69	50	34%	64%
4	38	40	60%	48%
5	40	57	81%	53%
6	68	50	46%	68%
7	31	56	100%	98%
8	64	49	35%	38%

Key: S ï Stable, **T** ï Tippy, **1st** ï Start up, **SS** ï Steady State

- Up to 49% of capacity (green)
- Up to 79% of capacity (amber)
- Up to 80% and above capacity (red lining)

Table 14 - Table showing the calculate P:C Ratio over Kerb terrain in the stable and tippy RAP

4.6.5 PC ratio - First push over all the terrains

Figure 19 shows the average P:C Ratio over all terrains during the Start up phase with the RAP in the stable and tippy position. On average subjects do not exceed their capacity to perform during the start up phase of the push. However, when looking at the range of P:C Ratios measured (see error bars) it can be seen that there are certain individuals are more likely to reach and exceed their P:C Ratio.

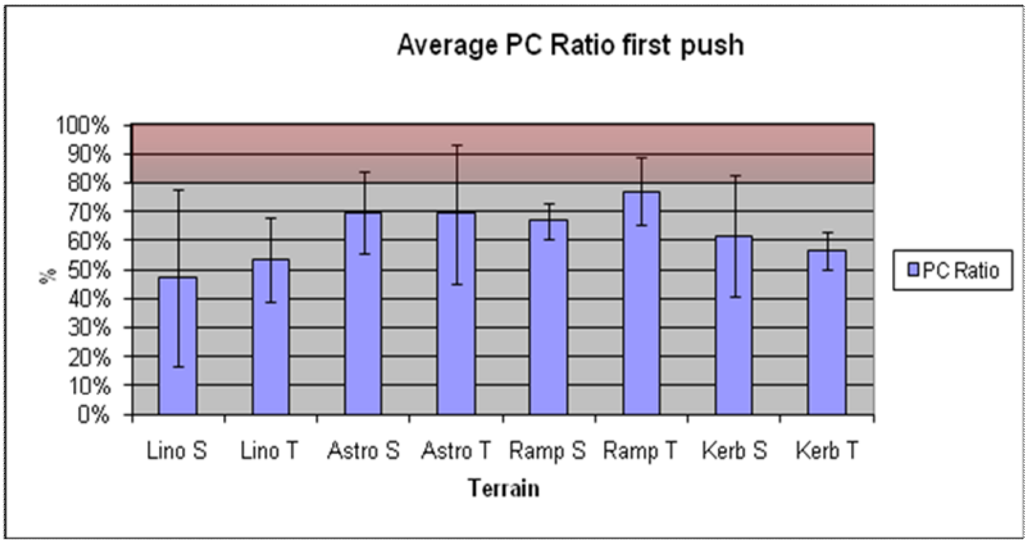


Figure 19 - Graph showing the average P:C Ratio over all terrains during the Start up phase with the RAP in the stable and tippy position

4.6.6 PC ratio - steady state over all the terrains

Figure 20 shows the average P:C Ratio over all terrains during the Steady State Phase with the RAP in the stable and tippy position. On average it can be seen that the readings are lower than those recoded in the start up phase of the push, and again when looking at individual cases (see error bars) it can be seen that there are no individuals who ŕed-lineô in steady state performance.

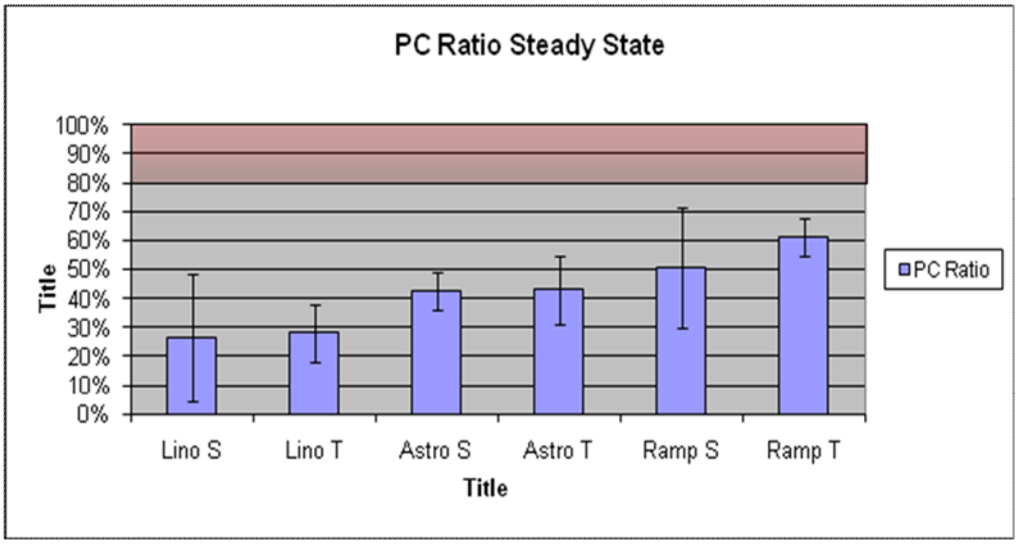


Figure 20 - Graph showing the average P:C Ratio over all terrains during the Steady State Phase with the RAP in the stable and tippy position

Looking at the difference with the two types of pushes there is an overall increased P:C ratio during the first push in comparison to the steady state push. The average P:C ratio is higher on the ramp than on all other terrains. First push is 67 % in the stable set up compared to 51% in the steady state. In the tippy set up the ratio is 77% on first push compared to 61% at steady state. When performing in the more tippy set up subjects have an overall increased PC ratio when performing on a ramp. This is the case for all the terrains apart from the kerb which shows a reduced PC ratio at first push when performing in the tippy set up.

5 Discussion

5.1 *Rear Axle Position*

As highlighted in the Literature Review (2.8), most clinicians currently assume that moving the rear axle towards the front of the wheelchair will make the wheelchair significantly easier to push by reducing the weight through the front castors, as suggested by Brubaker (Brubaker 1986). A 4.4cm forward adjustment of the RAP has been shown, theoretically, to give an 18% reduction in rolling resistance (Tomlinson 2000) for static conditions and a stationary trunk. Tomlinson's theoretical approach is supported by subjective reports of wheelchair users; who find a wheelchair easier to propel when it is adjusted in a 'tippy' setup. This suggests that a reduction in rolling resistance should result in reduced propulsion forces, something that is widely assumed in the available literature (Brubaker 1986; Tomlinson 2000; L. H. V. van der Woude et al. 2003). However, this study does not support this assumption, as although castor weight is indeed reduced in a less stable setup, this does not directly translate to reduced propulsion forces. These results were surprising and contrary to current clinical practice, which suggests that setting up a wheelchair in a less stable configuration assists in reducing cadence and push rim forces, thus achieving optimal propulsion efficiency (M. L. Boninger et al. 2000). This change could be due to the postural adaptations of the wheelchair users, which occurs during the dynamics of wheelchair propulsion.

5.2 *Effect of trunk flexion*

Kirby demonstrated the extent that the leaning of the wheelchair occupant had on wheelchair stability, but determined this only statically (RL Kirby et al. 1995). We are not aware of any studies to date that have fully examined what happens dynamically, when there is a cyclic shift in posture to maintain stability. The present study has adopted equations developed by Tomlinson (2000), described below, to evaluate the shift in the centre of mass in the x direction, using data generated in this study.

$$fc = mg\frac{x}{wb}$$

Equation 4: Castor Force Equation, where fc is the castor force (both castors combined), m is the mass (wheelchair plus occupant), g = acceleration due to gravity, x = horizontal distance from the rear axle to the centre of mass of the system, wb = wheelbase

The average centre of mass position can be viewed as a measure of how far forward, and backward, the user is leaning, with the subject initially moving towards the rear axle at prime position and moving forwards until a maximum castor force is reached.

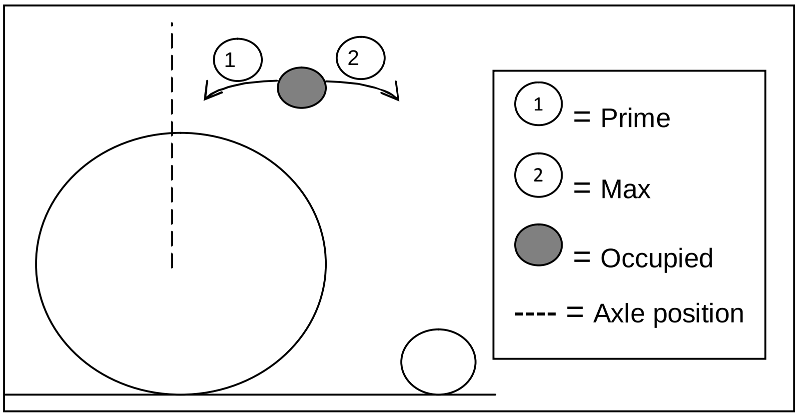


Figure 21: Demonstrating Centre of Mass position during propulsion of a wheelchair (occupied refers to the user sitting in the wheelchair).

This periodic trunk motion may cause a torque during the propulsion phase, promoting an increase in propulsion efficiency, as the acceleration of a mass going forward is the equivalent of the application of a force (L. H. V. van der Woude et al. 1989). This could explain why, although the castors become more loaded, this does not translate to increased propulsion force.

If Equation 4 is applied, when pushing on Lino, subjects lent back approximately 1cm (7.3cm ï 6.3cm in the stable set up, 4.5 ï 3.6 in the tippy set up) to prepare for prime, but when on Astro they remained virtually stationary (Figure 21 & Figure 22). As the Astro offers greater rolling resistance, it might have been expected that participants may take advantage of the grip offered and lean back more, which would have given them a bigger stroke angle. It would also offer the opportunity to flip the castors to overcome the resistance. However, the research showed this not to be the case, as stroke angle remained constant for all terrains.

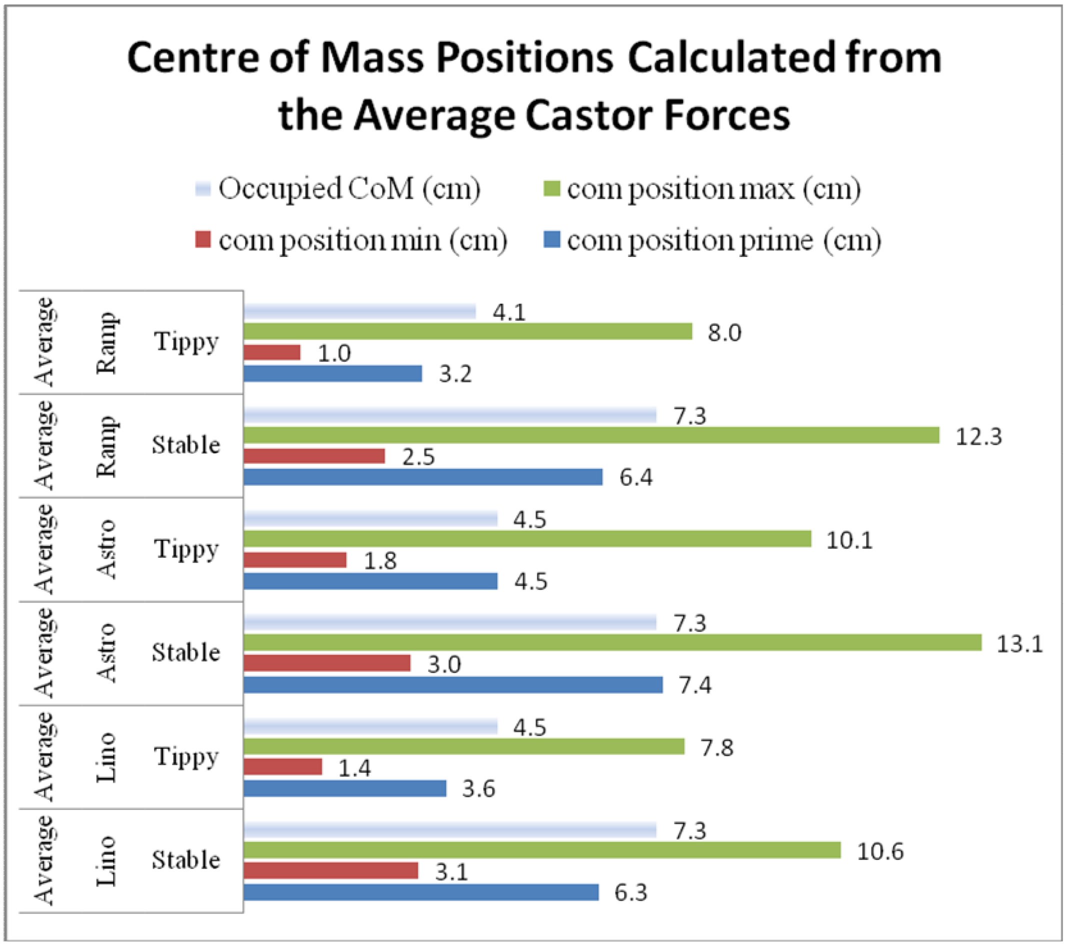


Figure 22: Centre of Mass positions calculated from the average castor forces

Secondly, the amount of trunk movement in the forwards direction (x direction) during the propulsion cycle (i.e. max position minus prime position) is relatively constant between tippy and stable set-ups for both Astro (5.7cm for stable and 5.6cm for tippy) and lino (4.3 cm for stable and 4.2cm for tippy). This supports the findings that stroke angle is not significant between the different set-ups. It would however suggest that occupants have a certain range of stroke angle, regardless of where they start on the rim.

The max COM position for the ramp terrain is found to be less than that during the Astro condition for both Tippy and Stable setups. This is most likely due to the fact that the horizontal position of the centre of mass is calculated from the castor forces, which are affected by gravity. Gravity could therefore decrease the loading on the castors, perhaps explaining why the x position appears to be less than that found on the Astro.

5.3 Effect of terrains on Propulsion Forces

As expected, the study found that greater propulsion force is needed when pushing the wheelchair on Astro and up a ramp, but very little difference could be seen between the Astro and the ramp terrains. It could therefore be assumed that the increased propulsion force is due to the necessary extra force required by the wheelchair user to overcome the effect of the gravity and the subsequent greater rolling resistance on propulsion, a view supported by (M. L. Boninger et al. 2000; DiGiovine 2006). For example, on the analysis of push two, the results show a greater Peak Mz on the ramp (34.6Nm) compared to the lino (20.5Nm). This is representative of all the pushes analysed (Table 6).

Koontzôs (2005) study identified that the start up phase of the push generated greater propulsion force (Push 1 ï 25.2Nm) and Push 2 (22.6Nm) compared to steady state (13.4Nm) on concrete surfaces and that these forces were greater on those surfaces imposing greater resistance (A. M. Koontz et al. 2005). In comparison, Cowanôs results from a tile surface show ñlower forces for strokes 1 and two, but higher for the steady state, at (89+/-27.1 N)ò(Cowan et al. 2008).

However, although the present study supports the findings of greater propulsion forces during the start up phase, there were no significant differences between the other terrains analysed. Another consideration is that when performing on the terrains imposing greater resistance, velocity was reduced with cadence remaining at similar levels (Table 6), this could indicate that subjects did not push faster or more frequently to overcome the changes in the terrains.

5.4 RAP and the effect on ascending a kerb

The kerb analysis demonstrated that a greater first propulsion force was typically needed with the RAP rearwards (stable), compared to the less stable configuration. For a clinician this would make sense, as when teaching users to flip their castors, it is much easier for them to achieve this with a less stable configuration. This also supports Kauzlarich & Collins (1988) view that, excess rear stability can limit a wheelchair usersô ability to lift their front castors. It is pertinent to note here that the Peak Mz recordings during the kerb task were not dissimilar to those readings on other terrains. This indicates that kerbs are not necessarily more effortful than the other terrains for the participants, rather perhaps, that it is skill and technique which are most influential, a view supported by Hashizume when exploring wheelchair users accessing gaps and steps between trains and platforms (Hashizume et al. 2007). It would be interesting to repeat such a test with less experienced wheelchair users.

5.5 P:C RATIO

Although recommendations have been made by the Paralysed Veterans Clinical Guidelines (Paralyzed Veterans of America 2005) on minimising frequency and force during wheelchair propulsion, there is no known research to date that identifies absolute force or push frequencies linked to the development or prevention of upper limb dysfunction. Until thresholds are identified, clinicians might focus on methods of identifying force and reducing the amount of force generated during propulsion, which will undoubtedly help them look at individual capabilities. It is also important to assess what is acceptable for a specific user, supporting the application of the ICF (International Classification of Functioning, Disability and Health 2001) in terms of ñperformanceò and ñdemandò. This is where the P:C ratio can be used, as a lowering of the P:C Ratio will always be beneficial, regardless of where the ñred-liningò limit is set.

The study calculated the P:C Ratio of each individual for each terrain, as outlined in WOWSUP 2006 (Nicholson,G et al, 2006). The results suggest that, as the terrain resistance increases, there is an increased occurrence of subjects approaching and exceeding their capacity. This was seen mostly when subjects perform in the stable set up.

The results also suggest that whilst there is an increased frequency of óred-liningô occurring in the more stable set up, subjects perform more frequently within the amber range of their capacity (50-79%) in the ótippyô set- up. It is not known if less upper limb injuries would occur if users exceeded their capacity on some terrains but less frequently, or whether performing just below their capacity more regularly might increase injury rates. Such questions continue to pose a challenge for the clinician when advising wheelchair users on how to minimise the risk of injury to upper limbs during functional mobility and whether a risk assessment is indicated.

There was an increased incidence of subjects reaching and exceeding their P:C Ratio when performing on the ramp in the ótippyô set up. This is a different pattern to that seen on both level terrains, which show an increase incidence of óred-liningô in the more stable set up. The ramp results also demonstrates an increased effort in the steady state part of the push, something which was not seen in other terrains. This could be related to the constant effort required to complete the task of ascending a ramp and that, when on other terrains, subjects can coast the later stages of the push and consequently reduce the need to push as hard or as frequently. WOWSUP (2006) reported a similar pattern of results to this effect. When ascending a ramp, the results demonstrated a series of what looked like, ófirst pushesô throughout the task, with each push showing similarities. WOWSUP also observed a consistently higher P:C Ratio when users ascended a slope.

The SmartWheel data supports these findings as there is a greater Peak Mz when subjects perform on a ramp. It could therefore be considered that regular propulsion on such terrain may further increase the likelihood of users developing upper limb injuries, due to the increased and cumulative loading on the arms, a view also supported Mercer (J.L Mercer et al. 2006). It is therefore very important that wheelchair users who regularly perform on such terrains, explore ways to reduce stress on their upper limbs and preserve function, something which is advocated in Riceõs study (Rice et al. 2008).

There were two subjects (5&7) whose results were particularly interesting in relation to their P:C Ratio. Both subjects consistently appear to find performing the tasks more difficult in the more stable set up. Both of these subjects have higher level injuries (along with Subject 2). In considering this, it maybe that it is less likely for these subjects to exceed their capacity to perform when the RAP is positioned in a more tippy set up. One may consider that subjects with higher level spinal cord injuries require a more ótippyô set up in order to reduce the likelihood of óred liningô and additionally as outlined by Boninger, may reduce the prevalence of upper limb injuries, although a causative relationship has not been demonstrated (Boninger 2000). The ramp performance results challenge this view as on the ramp, subject 7 red lined in a more tippy set up and not in the stable set up. One possible reason for this might be that users with a higher level injury lack postural control. This results in an óall or nothingô trunk flexion, and consequently, they shift their centre of mass further forward, working harder to prevent tipping rearwards when on a ramped surface. The same subjects (5 and 7) can consistently be seen to exceed their P:C. ratio over all terrains and again, both subjects have spinal cord injuries in the upper thoracic region. Although the study carried out no analysis on injury level, it could be considered that subjects with higher level SCI might be more likely to exceed their capacity to perform, especially on the ramp.

Another interesting outcome was seen on the kerb terrain with subject 2. This user consistently performed all functional mobility tasks well within their capacity to perform, but exceeded their capacity when performing the kerb. They were also the only subject unable to perform the kerb task with the RAP in the ótippyô set up. This study demonstrates that there is a need for clinicians to consider a wide range of functional mobility tasks when examining capacity and that it is skill and individual technique which will also affect the outcome of tasks.

The P:C Ratio measurement, an example of an isometric strength test, could be further developed and refined for clinical practice. Sabick *et al.* (2004) discussed that, because the maximal force produced by a muscle varies with both joint angle and velocity, the ideal method for determining capacity at any given time during the propulsion cycle would take both these variables into account. We should consider that whilst Sabick is correct, a userõs performance and capacity can also be influenced by the set-up of their wheelchair. Furthermore the development of a ñdynamicò P:C ratio would require a clinical practical method to measure joint angles and full kinematics, such as that used in gait analysis. The everyday practice of fitting a wheelchair has a long way to go before this level of sophistication could be proposed as ña best practiceò method. This project was established to develop a simple way of estimating the capacity and performance of wheelchair users in real life situations and, by using the Smart Wheel, the variables outlined by Sabik were incorporated. Furthermore, the study found that both cadence and stroke angle are largely insignificant across set-ups and terrains.

The P:C Ratio is a test that could be carried out in clinical practice, with a SmartWheel or similar instrumentation, to determine the performance limitations of the wheelchair user. It might then be possible for adjustments to be made to the wheelchair and for a re-testing cycle to be implemented in much the same way clinicians use pressure mapping to evaluate cushions. Ultimately, the aim would be to achieve greater levels of mobility without reaching the redline zone, thus reducing the risk of injury.

Further criticism in the application of this measure is the subjective nature of the evidence to support the 80% threshold which was chosen by Nicholson (2006) as the level where an individual would exceed their capacity and óred lineô This was done by subjective questioning. Although the application of this measure is easy for clinicians to understand there is no evidence offered linking this threshold to actual over-use injury associated with wheelchair propulsion. It is also important to recognise that wheelchair propulsion is only one activity undertaken by people with SCI that has the potential to injure the structures of the upper extremities. Transfers are also thought to contribute substantially to this risk. As a robust óevidence-basedô clinical model will be elusive, with a degree of common-sense and clinical judgement needed the application of The Borg 15 point scale may add more value to such a system. Also in applying this scale it must be observed that some individuals may perform consistently just below the 80% level. It would be wrong to assume that individuals who perform in the Amber range of the scale are not at risk. Therefore careful interpretation is required of the results to promote an accurate picture of individual performance.

Since completion of this study, research has been published by Cowan (2008) who uses Peak Force and velocity as a measure.

However, the limitation of such engineering parameters is that they do not put the performance of the user into context of their capacity, therefore the study is unable to identify whether users were comfortable or not in completing these tasks.

The results demonstrate that there are some users who indicate a P:C Ratio of over 100% (Table 11, Table 12, Table 13, Table 14). It is assumed that in generating this data that an individual is greatly exceeding their own P:C Ratio and is therefore in an excessively dangerous state. However, it is believed that since capacity is measured isometrically the dynamic use of the trunk and the alteration in posture during active propulsion increases actual capacity. The capacity measurement can be improved by changing the protocol for capacity by asking the user to accelerate as hard as possible and obtaining the highest push peak force during that task. One point to note, is that such a measure is personal to each individual and when working with people, situations do change. A measurement can only be taken with a particular person on a particular day therefore results may vary depending on their physical/mental/emotional state. This is a refinement for future consideration.

It is recognised that whilst the SmartWheel remains the most obvious tool to use for generating data on propulsion it is not accessible for many clinicians in practice. Cowan (2008) recognises that clinicians without access to a Smart wheel could use push frequency and velocity without such tools. However, this would not provide the required data to calculate the P:C Ratio. Further investigation is required to examine what other methods are available to clinicians to generate the data required to apply such a scale.

5.6 Other Smart Wheel Parameters

Some of the additional data recorded by the SmartWheel indicates that Velocity and Peak Average Force do provide significant results when the RAP is adjusted.

5.6.1 Velocity

Literature supports the view that manual wheelchair users should be able to achieve a minimal velocity for function and performance regardless of their diagnosis and that velocity, as a measure, should not be underrated (Cowan et al. 2008). It could be argued that the subjects may perform faster on certain terrains with the RAP forwards, but this is not supported by the velocity data, in reality, most performed slightly faster in the more stable set-up. The research allowed participants to self-select their speed, as recommended by the SmartWheel protocol.

The difference in velocity is minimal between the tippy and stable set-up. Cowan's study used 1.06 m/s as their threshold velocity (Cowan et al. 2008). This is viewed as the average walking velocity required to safely cross an intersection (Hoxie & Rubenstein 1994) . In this study, as expected, this level was not achieved for push one but was achieved and exceeded on the level terrain. Users performing on the ramp, did on average, fall below this threshold as did some subjects on the Astro terrain, especially in the tippy set up. However, it could be argued that wheelchair users would not be expected to perform at the same speed over more challenging terrains, much like in gait, where speed would be different walking on the flat compared to walking uphill. Although Cowan does not reflect on this point, she recommends that for those individuals who fall below this velocity, a programme should be designed to help achieve a threshold velocity (Cowan et al. 2008). Such interventions may include strength training, review of wheelchair setup or alternative mobility, such as powered provision. Also, what should be questioned is the regularity of negotiating such terrains as part of their everyday life, to determine how realistic such interventions are. For example, should a subject not have the need to negotiate a ramp regularly then training in this area is not relevant to them. The average velocity for each surface was calculated and indicated that participants pushed at a slower average speed on the ramp and Astro terrains (1.0 m/s) than on the lino (1.2 m/s). Similar results were found by Koontz et al and Cowan, who summarised that, users selected a lower velocity as the surface difficulty increased (Cowan et al. 2008; A. M. Koontz et al. 2005). It is important to consider that subjects in all studies self selected the speed they were travelling at for all the terrains so perhaps they could have gone faster if they had wanted to. However, it was felt to be difficult to set specific speeds due to the difference in presenting ability and that the subjects ‘usual’ technique may not be represented.

Cowan also reported an increase in push force but with the same push frequency and stroke length on these more difficult terrains, something which is partly recognised in the study. Whilst terrain does indeed influence push force, cadence or stroke angle is not significant (Cowan et al. 2008).

Cowan also outlines that clinicians need to attempt to preserve velocity, while minimising force and push frequency, to help delay upper limb pain and dysfunction (Cowan et al. 2008). Increased velocity is related to increased pushing forces, however, this study did not see a correlation between increased velocity and an increase in propulsion forces.

5.6.2 Peak Average Force

The results from the present study indicated that subjects had a lower peak average force when the rear axle was in its most stable position (<0.047). It is known from the SmartWheel user handbook, that a lower ratio indicated that the peak average force is closer to the average force demonstrating a smoother push (SmartWheel Users Guide 2005) and that a smoother push is thought to indicate a lower risk of upper limb injury (Paralyzed Veterans of America 2005). The same report recommends adjusting the rear axle as far forwards as possible (Paralyzed Veterans of America 2005). However, given the findings of this study the two pieces of advice are conflicting; as a more forward RAP led to a less smooth push.

6 Conclusion

The impact of the rear axle position on stability is easy to understand, in that the more forwards the rear axle, the more unstable the wheelchair is. However, it is more difficult to anticipate the effect that such adjustments have on straight line functional mobility. The study shows that when the RAP is set forwards, in a tippy position, the castor forces are reduced and that terrains generating higher rolling resistances caused the greatest increase in the castor forces. However, increased castor forces were not directly matched by an increase in push-rim forces, moving away from current theory (Brubaker 1986; Tomlinson 2000). It had been anticipated that users would push harder or more frequently in a more stable set up, but, after the first push the velocity of the wheelchair was not affected by the RAP and the cadence was also unaffected for all terrains. Users were not found to push harder or more frequently in either set up, but actually travelled slower in the tippy set up.

Subjects were found to dynamically adjusted their trunk posture during tasks. This resulted in castor forces showing no difference between level ground and the ramped terrains. It is likely the forward postural change (which adds load to the castors) and the gradient of the slope (which naturally unloads the castors) were comparable and therefore cancelled the overall effects of each other.

The new measure, the P:C Ratio, indicated that certain individuals are more likely to exceed their capacity to perform and the study highlighted that this maybe more indicative of a higher level spinal cord injury. In the stable set-up, as well as when rolling resistance was higher, the P:C Ratio was generally higher.

Further research would be required to fully understand the complexities of postural changes used intuitively by both experienced and inexperienced wheelchair users to assist in optimising wheelchair set up for optimum performance. From clinical experience, we can assume that an individual's level of injury and technique can profoundly affect the balance between stability and 'tipability' of the wheelchair. The study did not examine in detail the level of injury and the effects of posture during propulsion. This would require further analysis incorporating use of motion analysis such as CODA, a recommendation for future work.

Considerable data would be required to determine at what level the P:C Ratio should be set for red-lining, so that it correlates with the chances of upper limb injury presenting at a later date. In the short term, the principles of this measure could be used by clinicians to assess relative adjustments to the wheelchair set-up, in order to optimise performance or suggest alternative methods and techniques. It is important that users are educated, especially those who are less experienced, about these ratios and provided with appropriate advice about the safe upper limits of exertion in order to minimise the risk of upper limb injury. Such measures could prove to be of great value in reducing such injuries.

The P:C ratio would appear to be an ideal tool for many fields of rehabilitation. It could be considered in the post injury stage to provide education to first time wheelchair users as a preventative measure and as an education tool outlining the effects of performing on different terrains. It could also be used during the rehabilitation of injured wheelchair athletes, as a method of avoiding over exertion during the rehabilitation process, facilitating recovery and assisting in prescribing an individual's rehabilitation programme. It could be viewed as a similar tool to pressure mapping, which is currently used to assist with cushion evaluation and prescription. The P:C Ratio is certainly a tool in its infancy, but could assist in identifying risk and allowing wheelchair users and clinicians to make informed decisions around functional mobility. Although these benefits can be seen in its application, the data generation is perhaps too complicated for general clinical practice. The equipment is also expensive and requires time to analyse. The new SmartWheel software should overcome the time element of this process, but funding will remain a key issue, particularly in today's NHS climate.

How a wheelchair is set up in relation to the user and the physical status of the user themselves, are critical factors to determine how a person is able to access their environment. For wheelchair prescriptions to be effectively completed and for wheelchairs to be optimally set up, it is essential that clinicians develop greater skills and an in-depth understanding of functional wheelchair use. This applies not only to the active wheelchair users, but for all self propelling wheelchair users. It is essential that any knowledge gained is communicated to those at the forefront of clinical practice in order to promote best practice. Based on the results of this study a document, 'Key points for Clinicians' has been created (Appendix 5). The overriding summary is that, with appropriate skills training and experience, a wheelchair user can learn to compensate for the changes in castor loading. Therefore, optimisation of wheelchair set-up is not always a scientific measure but a clinical compromise according to an individual's skill and needs within their environment.

6.1 Limitations and recommendations for Future Work

There are three major limitations of the current study. The first being a lack of manoeuvrability and turning tasks within the protocol. This study was unable to look at this, due to the limited rotational ability of the instrumented castor and difficulties produced when the wheelchair was not propelled in a straight line. Could manoeuvrability have been addressed in the data gathering process, then this may have demonstrated a change in propulsion forces with the RAP in its most forwards position. A more sophisticated linear bearing with a separable cage could possibly be used to overcome this difficulty in future studies. Cost and time constraints would not allow it to be included in this study.

The second limitation was the number of participants recruited for the study. Although it was thought 8 people would be enough to allow for statistical inferences to be made from the results, this was unfortunately not the case.

Thirdly, due to the complexities of the equipment configuration and electronics, the study could only access one manual wheelchair for testing, a 17inch Quickie GPV. Clinically it is agreed that to fully optimise a chair then the prescriber needs to select a chair of an appropriate width. However, it should be noted that most of the subjects who participated within the study were around 17 inches wide, and that the wheelchair was modifiable in other ways to optimise individual set up as much as possible. There were only minimal adjustments to the chair required throughout the testing. It could also be considered that use of only one wheelchair type may have had an impact on the results. The GPV has very specific axle adjustment which is limited to a certain range and it was the extremes of this range that were tested. Data only represented the extremes of the most stable and the most tippy and this may reflect some of the results seen. Selection of an alternative lightweight wheelchair may have shown a different set of results.

On a more general note a lack of time and money prevented a large and varied sample population. In particular, those without a SCI and those with a high level injury (affecting upper limb hand function) were excluded from the study. Newsam et al (1996) suggests that people with higher lesions must work near or at their maximum capability for basic community function and perform at an increased velocity than lower thoracic injuries. However, a limitation to this study was that users with a higher lesion could not be tested due to the inaccuracy of the readings from the SmartWheel. It was also observed that the majority of subjects at this level show a preference to using the tyre rather than the rim. With appropriate measures, this could be an area of future investigation, focussing on the efficiency between tyre and rim propulsion. Another interesting aspect with higher lesions is the reduced capacity to use dynamic postures, due to the lack of trunk control, and seeing less of a shift in their Centre of Mass (Majaess 1992), something which would benefit from further investigation. The exclusion of such users within this study is unfortunate, as this group could be defined as one of the most at risk of upper limb injury. From clinical experience, they are users that commonly require fine tuning to their wheelchair set up, in order to achieve optimal efficiency and performance. Therefore it is imperative that alternative measurement options be explored in order to understand this user group further.

There are many interesting discussion points that have arisen from this study and are worthy of future research. In particular, the complex interplay between rolling resistance and its relationship with the castor loading, during functional tasks would benefit from a higher sample size and also from measuring the kinematics of the upper body. The impact of RAP on the kinematics would also be of interest.

Further work is needed to develop a clinically relevant tool to predict upper limb injury. Although this study has taken step towards this by implementing the P:C Ratio for the various tasks and set-ups used.

There is still a lot of work to be done to make it a functional clinical tool. This could be accomplished by developing a link between the maximum contractions of various upper limb muscle groups and the body's physiological reactions to exertion.

The analysis was restricted to all subjects who had a SCI. Although this proves beneficial when comparing with other studies, this is not reflective of the manual wheelchair population. It should be noted that the proposed conclusions were based on a small number of subjects and that there was a high level of variation found among the subjects; warranting that these results be replicated (to ensure the generalisation of the trends observed) before using the results to assist other groups.

From this study, we can see that when the RAP is set forwards (tippy) on an experienced user's wheelchair it does not necessarily translate into lower propulsion forces, as has (until now) been widely believed. However, care should of course be taken in translating any laboratory test results into wheelchair performance within functional environments. Additionally, these tests were performed for straight line activities and, apart from that of the kerb, did not therefore include the influence of real-life manoeuvrability tasks.

7 References

The web addresses cited were current in November 2009

Audit Commission, 2000. *Fully Equipped - the provision of equipment to lder or disabled people by the NHS and social services in England and Wales*,

Barnard.T. 2006 - (http://personal.health.usf.) http://personal.health.usf.edu/tbernard/HollowHills/SMWDesignM20.pdf)

Boninger, M.L. et al., 2000. Manual wheelchair pushrim biomechanics and axle position. *Archives of Physical Medicine and Rehabilitation*, 81(5), 608-613.

Boninger, M.L. et al., 1999. Wheelchair pushrim kinetics: Body weight and median nerve function. *Archives of Physical Medicine and Rehabilitation*, 80(8), 910-915.

Borg Scale 2009 http://www.brianmac.co.uk/borgscale.htm

Brose, S.W. et al., 2008. Shoulder ultrasound abnormalities, physical examination findings, and pain in manual wheelchair users with spinal cord injury. *Archives of Physical Medicine and Rehabilitation*, 89(11), 2086-2093.

Brubaker, C.E., 1986. Wheelchair prescription: an analysis of factors that affect mobility and performance. *J Rehabil Res Dev*, 23(4), 19-26.

Burning, M. & Schmeiler, M., 1999. *Manual Wheelchairs: Set up and Propulsion Biomechanics*, UPMC Health System.

Calder, C.J. & Kirby, R.L., 1990. Fatal Wheelchair-related Accidents in the United States. *American Journal of Physical Medicine & Rehabilitation*, 69(4), 184.

Collinger, J.L. et al., 2008. Shoulder biomechanics during the push phase of wheelchair propulsion: a multisite study of persons with paraplegia. *Archives of Physical Medicine and Rehabilitation*, 89(4), 667-676.

Cooper, R.A., Stewart, K.J. & VanSickle, D.P., 1994. Evaluation of methods for determining rearward static stability of manual wheelchairs. *J Rehabil Res Dev*, 31(2), 144-7.

Cooper, R. *et al*., 1997. Methods for determining three-dimensional wheelchair pushrim forces and moments: A technical note. *Journal of Rehabilitation Research and Development*, 34(2), 162-170.

Cowan, R.E. et al., 2008. Preliminary outcomes of the SmartWheel Users' Group database: a proposed framework for clinicians to objectively evaluate manual wheelchair propulsion. *Arch Phys Med Rehabil*, 89(2), 260-8.

De Groot, S. et al., 2002. Wheelchair propulsion technique and mechanical efficiency after 3 wk of practice. *Medicine & Science in Sports & Exercise*, 34(5), 756.

DiGiovine, C.P., 2006. Advances in Manual Wheelchair Technology. *Topics in Spinal Cord Injury Rehabilitation*, 11(4), 1-14.

DiGiovine, C.P. et al., 1997. Determination of the bilateral pushrim moments about wheel axle and work required to transverse obstacles. In pp. 20ï 24.

van Drongelen, S. et al., 2005. Mechanical Load on the Upper Extremity During Wheelchair Activities. *Archives of Physical Medicine and Rehabilitation*, 86(6), 1214-1220.

Gellmen H, Sie IH & Walters RL, 1988. Late complications of the weight baring upper extremity in the paraplegic patient. *Clinical Orthop*, 132-135.

Hashizume, T. et al., 2007. Study on the accessibility and assistive device between platforms and trains for manual wheelchair user. In *SICE, 2007 Annual Conference*. pp. 494-498.

Hoxie, R. & Rubenstein, L., 1994. Are older pedestrians allowed enough time to cross intersections safely? *Journal of the American Geriatrics Society*, 42(3), 241-244.

Kauzlarich JJ, Collins TJ. 1998. Performing a wheelchair wheelie balance. In: de Groot G, Hollander AP, Huijing PA, van Ingen, Schenau GJ, editors. International series on biomechanics, *BIOMECHANICS*. XI-A. Amsterdam: Free University Pr; 1998. p 507-12.

Kirby, R.L., 1996. Wheelchair stability: important, measurable and modifiable. *Technology and Disability*, 5, 75-80.

Kirby, R.L. et al., 1996a. Adding loads to occupied wheelchairs: Effect on static rear and forward stability. *Archives of Physical Medicine and Rehabilitation*, 77(2), 183-186.

Kirby, R.L. et al., 1996b. Adding loads to occupied wheelchairs: Effect on static rear and forward stability. *Archives of Physical Medicine and Rehabilitation*, 77(2), 183-186.

Kirby, R.L., Atkinson, S.M. & MacKay, E.A., 1989. Static and dynamic forward stability of occupied wheelchairs: influence of elevated footrests and forward stabilizers. *Arch Phys Med Rehabil*, 70(9), 681-6.

Kirby, R.L. & Dupuis, D.J., 1999. Clinical measurement of the static rear stability of occupied wheelchairs. *Archives of Physical Medicine and Rehabilitation*, 80(2), 199-205.

Kirby, R.L., McLean, A.D. & Eastwood, B.J., 1992. Influence of caster diameter on the static and dynamic forward stability of occupied wheelchairs. *Arch Phys Med Rehabil*, 73(1), 73-7.

Kirby, R.L. et al., 1994. Wheelchair stability and maneuverability: effect of varying the horizontal and vertical position of a rear-antitip device. *Arch Phys Med Rehabil*, 75(5), 525-34.

Kirby, R. et al., 1995. Wheelchair stability: effect of body position. *Journal of rehabilitation research and development*, 32(4), 367-372.

Kirkwood, B. & Sterne, J., 2003. *Essential Medical Statistics*, Blackwell Publishers.

Koontz, A.M. et al., 2005. A kinetic analysis of manual wheelchair propulsion during start-up on select indoor and outdoor surfaces. *J Rehabil Res Dev*, 42(4), 447-58.

Kotajarvi, B.R. et al., 2004. The effect of seat position on wheelchair propulsion biomechanics. *J Rehabil Res Dev*, 41(3B), 403-14.

Kwarciak, A.M. et al., 2009. Redefining the manual wheelchair stroke cycle: identification and impact of nonpropulsive pushrim contact. *Archives of Physical Medicine and Rehabilitation*, 90(1), 20-26.

Lachmann, S., Marks, L. & McCarthy, G., 1995. The provision of wheelchairs and special seating. Guidance for purchasers and providers. Working Group of the Royal College of physicians. *Journal of the Royal College of Physicians of London*, 29(3), 194-5.

Lal, S, 1998. Premature degenerative shoulder changes in spinal cord injury patients. *Spinal Cord*, 36, 186-189.

Leibel, G. & Patrick, L., 1998. *Recipe for a Wheelchair Prescription: Spina Bifida in the New Millenia*, Bloorview MacMillan Childrens Centre.

Loane, T. & Kirby, R., 1985. Static rear stability of conventional and lightweight variable-Axle-position wheelchairs. *Archives of physical medicine and rehabilitation*, 66(3), 174-176.

Majaess, G. et al., 1993. Influence of seat position on the static and dynamic forward and rear stability of occupied wheelchairs. *Archives of physical medicine and rehabilitation*, 74(9), 977-982.

Masse, L., Lamontagne, M. & O'Rain, M., 1992. Biomechanical analysis of wheelchair propulsion for various seating positions. *Journal of rehabilitation research and development*, 29(3), 12-28.

Medicines and Healthcare Regulatory Advisory , 2004. Guidance on the Stability of Wheelchairs.

Mercer, J.L. et al., 2006. Shoulder joint kinetics and pathology in manual wheelchair users. *Clinical Biomechanics (Bristol, Avon)*, 21(8), 781-789.

Newsam, C.J. et al., 1996. Temporal-Spatial Characteristics of Wheelchair Propulsion: Effects of Level of Spinal Cord Injury, Terrain, and Propulsion Rate1. *American Journal of Physical Medicine & Rehabilitation*, 75(4), 292.

Nicholson,G et al., 2006. In *The Workshop for Optimisation of Wheelchair Selection and User Performance*. North West Regional Spinal Injuries Centre, Southport.

Paralyzed Veterans of America, 2005. Preservation of the Upper limb following Spinal Cord Injury: a clinical practice guideline for health care professionals.

Pentland WE & Twomey LT, 1991. The weight baring upper extremity in women with long term paraplegia. *Paraplegia*, 29, 521-530.

Pope , P., 2005. Dynamic Assessment of Dynamic Seating. In Exeter 11-15 April.

Rasmussen, J. et al., 2004. *The Role of Mechanics and Optimisation in Ergonomics*, DK-9220 Aalborg East, Denmark : The AnyBody Group, Institute of Mechanical Engineering, Aalborg University, Pontoppidanstraede 101.

Rentschler, A.J., 2002. Analysis of the ANSI/RESNA Wheelchair Standards: A Comparison Study of Five Different Types of Electric Powered Wheelchairs.

Rice, I. et al., 2008. Manual wheelchair stroke characteristics during an extended period of propulsion. *Spinal Cord*, 47(5), 413-417.

Richter, W.M., 2001. The effect of seat position on manual wheelchair propulsion biomechanics: a quasi-static model-based approach. *Medical Engineering and Physics*, 23(10), 707-712.

Richter, W.M., Chesney, D.A. & Axelson, P.W., 1999. Mechanical Efficiency of low-impact Pushrims. In RESNA Press.

Rose , L. & Ferguson-Pell , M., 2002. National survey of wheelchair provision to people with spinal cord injury (1). *Brit. J. Therapy and Rehab.*, 9(10).

Rose, L.S. & Ferguson-Pell, M.W., 2009. Wheelchair provision to people with Spinal Cord Injury before and after the Voucher Scheme. In *Wheelchair provision to people with Spinal Cord Injury before and after the Voucher Scheme*. Warwick University.

Sawatzky, B.J., Miller, W.C. & Denison, I., 2005. Measuring energy expenditure using heart rate to assess the effects of wheelchair tyre pressure. *Clinical Rehabilitation*, 19(2), 182-187.

Sie IH et al., 1992. Upper Extremity Pain in the postrehabilitation spinal cord injured patient. *Archives of Physical Medicine and Rehabilitation*, 73, 44-48.

SmartWheel Userœ Guide,2005, Rev 1.2

Last Updated: December 23, 2005

Spinal Cord Injury Peer Support. 2008. *Definition of a Wheelchair.* Apparelyzed.com/wheelchair.html

SPSS Inc, 233 S Wacker Dr, 11th Floor, Chicago, IL, 60606-6307

Stover, S.L., DeLisa, J.A. & Whiteneck, G.G., 1995. *Spinal Cord Injury: Clinical Outcomes from the Model Systems*, Aspen Publishers.

Three Rivers Holdings 2009 http://www.3rivers.com/swcustomercare.php.

Three Rivers Holding LLC. 1826 Broadway Rd, Site 43, Mesa, AZ 85202

Tomlinson, J.D., 2000. Managing Maneuverability and Rear Stability of Adjustable Manual Wheelchairs: An Update. *Physical Therapy*, 80(9), 904.

Trudel, G. et al., 1997. Effects of rear-wheel camber on wheelchair stability. *Archives of Physical Medicine and Rehabilitation*, 78(1), 78-81.

Unmat, S. & Kirby, R.L., 1994. Nonfatal wheelchair-related accidents reported to the National Electronic Injury Surveillance System. *Am J Phys Med Rehabil*, 73, 163-7.

Vanlandewijck, Y, Spaepen, A, Lysens, R 1994 Wheelchair propulsion efficiency: movement pattern adaptations to speed changes. Medicine and Science in Sports and Exercise. The Journal of the American College of Sports Medicine

Vanlandewijck, Y., Theisen, D. & Daly, D., 2001. Wheelchair Propulsion Biomechanics: Implications for Wheelchair Sports. *Sports Medicine*, 31(5), 339.

Veeger, H.E.J., Rozendaal, L.A. & van der Helm, F.C.T., 2002. Load on the shoulder in low intensity wheelchair propulsion. *Clinical Biomechanics*, 17(3), 211-218.

Walsh, C.M., Marchiori, G.E. & Steadward, R.D., 1986. Effect of seat position on maximal linear velocity in wheelchair sprinting. *Can J Appl Sport Sci*, 11(4), 186-90.

West Midlands Rehabilitation Centre, 2005. Rehabilitation Engineers Handbook Stability Testing.

Wheatley, D.W., Berme, N. & Ferguson-Pell, M.W., 1980. A low-profile load transducer for monitoring movement during sleep. *Experimental Mechanics*, 20(6), 19-20.

van der Woude, L.H. et al., 1989. Seat height in handrim wheelchair propulsion. *Journal of Rehabilitation Research and Development*, 26(4), 31-50.

van der Woude, L.H.V. et al., 2003. Measurement of wheelchair rolling resistance with a handle bar push technique. *Journal of Medical Engineering & Technology*, 27(6), 249-258.

van der Woude, L.H.V., Veeger, H.E.J. & Rozendal, R.H., 1989. Propulsion technique in hand rim wheelchair ambulation. *Journal of Medical Engineering & Technology*, 13(1), 136-141.

Zatsiorsky, V., 2000. *Biomechanics in Sport: Performance Enhancement and Injury Prevention* 1st ed., Wiley-Blackwell.

8 Appendix 1 – Participant Information

8.1 Letter of Invite

ASPIRE Centre for Disability Sciences
Institute of Orthopaedics and Musculoskeletal Sciences
Royal Free and University College Medical School
Brockley Hill
Stanmore HA7 4LP
Tel: 0208 909 5471

Name
Address
Date

Dear

ACDS is a research team linked with the ASPIRE National Training Centre and University College London. We are currently working on establishing objective measures and new guidelines for the provision and optimisation of wheelchairs for people with a spinal cord injury.

We are currently looking for volunteers to participate in our research, with a particular focus on wheelchair stability. An information sheet has been enclosed and we will be contacting you by phone in the near future to establish if you are able to volunteer.

In the meantime if you have any further questions regarding the research please contact Lynne Hills or Zillah Bloomer at the ASPIRE Centre for Disability Sciences on Tel. 0208 909 5471.

Yours sincerely

Lynne Hills
Research Therapist

8.2 Participant Information Sheet

Royal National Orthopaedic Hospital NHS
NHS Trust

RNOH Stanmore
Brockley Hill
Stanmore
Middlesex
HA7 4LP

Tel: 020 8954 2300
www.rnoh-stanmore.org.uk

ROYAL FREE AND UNIVERSITY COLLEGE LONDON MEDICAL SCHOOL
ASPIRE CENTRE FOR DISABILITY SCIENCES
and
ROYAL NATIONAL ORTHOPAEDIC HOSPITAL TRUST
PARTICIPANT INFORMATION SHEET

The purpose of this consent form is to provide you with the information that you need to consider in deciding whether to participate in a research study which will enable wheelchair users to assess their propulsion ability, achieve greater levels of mobility with less risk of injury.

Study title: Dynamic Stability Testing

Purpose of Research Project

To maximise performance in a wheelchair, wheelchair set up, wheelchair skills and an effective propulsion technique are essential. Key factors are stability and pushability.

There have been many studies measuring the static stability of manual wheelchairs but very little on dynamic stability or how this relates to static stability. Dynamic or functional stability could be best described as how change in the weight distribution and centre of mass (COM) of the wheelchair and user affects rolling resistance and ótippinessô

Static weight distribution and ñtipò angle can be measured from a tilting weigh platform. This part of the research has been completed. We hope that by using instrumented castors and an instrumented handrim (SmartWheel^{TM}) it will be possible to gather data on castor weight (dynamic weight distribution), pushrim forces and chair acceleration and deceleration during each propulsion cycle. These will provide us with measurements to give clarity on the term stability.

We plan to attempt to measure dynamic stability of experienced wheelchair users during functional mobility tasks.

Procedure

We will ask you to propel a wheelchair that has been set up to test dynamic wheelchair stability. Although the control wheelchair is a set size it will be carefully setup for you to match your body build and level of spinal cord injury, as much as is achievable.

You will be asked to perform a number of functional mobility tasks which will include propulsion along a lino floor, Astro turf, a standard ramp and ascending a kerb. Each of the tasks will be performed with the axle in its most stable position and its most tippy position.

If you decide to take part in this study, you will be invited to attend a 2 hour assessment at the Royal National Orthopaedic Hospital.

There are no expected risks associated with this study, other than those normally associated with wheelchair use. This study may produce a direct benefit to you now as an individual, and it could benefit you and many other wheelchair users in the future, as well as helping clinicians understand how to provide assessments that can have the greatest benefit in terms of wheelchair prescription and set up.

All information about you obtained during the study will be kept in files that will be kept confidential.

Taking part in this study is completely up to you. You can refuse to take part or withdraw from the study at any time and such a decision will not affect you care in any way.

If you have any questions, you can reach Professor Ferguson-Pell at 0208 909 5471 or Lynne Hills on 0208 385 8111and they will do their best to answer them.

8.3 Consent sheet for participants

Royal National Orthopaedic Hospital
NHS Trust

RNOH Stanmore
Brockley Hill
Stanmore
Middlesex
HA7 4LP

Tel: 020 8954 2300
www.rnoh-stanmore.org.uk

CONSENT TO PARTICIPATE IN A RESEARCH STUDY

Title: Workshop for optimisation of wheelchair selection and user performance (Dynamic Wheelchair Stability)

I agree to take part in this study. I have read the Patient Information Sheet for this study and I understand what will be required of me if I take part in this study.

My concerns regarding this study have been answered by Professor Ferguson-Pell or his colleagues to my satisfaction. I understand that taking part is up to me and that I can withdraw from the study at any time without giving a reason and without affecting my normal care and management. I have read the above and agree to enter this research study.

Signing this form does not alter any of my legal rights. I have been informed of the procedure described above with its possible risks and benefits. I have been given a chance to ask any and all questions I have. I understand that, if I can think of more questions later, Professor Ferguson-Pell or his colleagues will answer them for me. I can reach them at 0208 909 5471. If these questions are not answered to my satisfaction I also understand that I may contact the secretary of the chairman of the Joint Research and Ethics Committee (020 909 5314) which has approved this study.

I understand that:

In case of emergency, the Royal National Orthopaedic Hospital Trust will give me emergency medical care if the medical staff of the hospital think it is needed. If care cannot be given at the Royal National Orthopaedic Hospital Trust, then Professor Ferguson-Pell or his colleagues will arrange for care by someone else. I also know that University College London or the Royal National Orthopaedic Hospital Trust has the right to stop the study at any time, or to drop me from the study.

I have received a copy of this form.

Participant.

Nameé é é é é é é é é é é é é é é é é .
Signatureé é é é é é é é é é é é é é é é
Dateé é é é é é é é é é é é é é é é é é

Investigator eliciting consent.

Nameé é é é é é é é é é é é é é é é é
Signatureé é é é é é é é é é é é é é é ..
Dateé é é é é é é é é é é é é é é é é ..

9 Appendix 2 - Castor Construction and Calibration

9.1.1 Instrumentation to measure Castor forces

In order to measure castor forces, force washers (Interface Inc, model LW2050-250 capacity 20lbf) were placed on the stem of each castor between the castor fork and the castor bearing. Initially, a rubber bung was placed between the nut and the bearing later to be replaced by a spring, as the bung was suspected to add hysteresis to the design. The spring allowed the castor stem assembly to be tightened while only gradually increasing the load on the force washer with each turn, consequently reducing the tendency for significant force increments to occur as the castor swivelled. Washers were used as spacers to accommodate the thread on the stem. Castor stems had to be specially fabricated from stainless steel to accommodate the thickness of the bearing and spring assembly and also maintain integrity as during the initial stages of testing, a prototype aluminium stem snapped.

Figure 23 - Castor construction

9.1.2 Calibration of the castor force transducer

Pre-calibrated, low-profile force plates used in the ergometer studies and described in detail by Wheatley et al (1980), were used as a benchmark to measure the force between the castors and the ground. Data was recorded using a Data Translation PCI A-D converter sampling at 200Hz (DT 332) and a specially developed program designed for the ergometer system using an Agilent Vee data acquisition system. While running the Tachyon system and the force plate system simultaneously, weight was progressively added to the wheelchair and then removed to produce a loading-unloading calibration. The test was then repeated with an occupant in the wheelchair transferring weight, first of all slowly by leaning forward and back, and then rapidly whilst propelling the wheelchair on the ergometer. In each case simultaneous measurements of actual load on the castor and measure load determined by the force washer could be obtained.

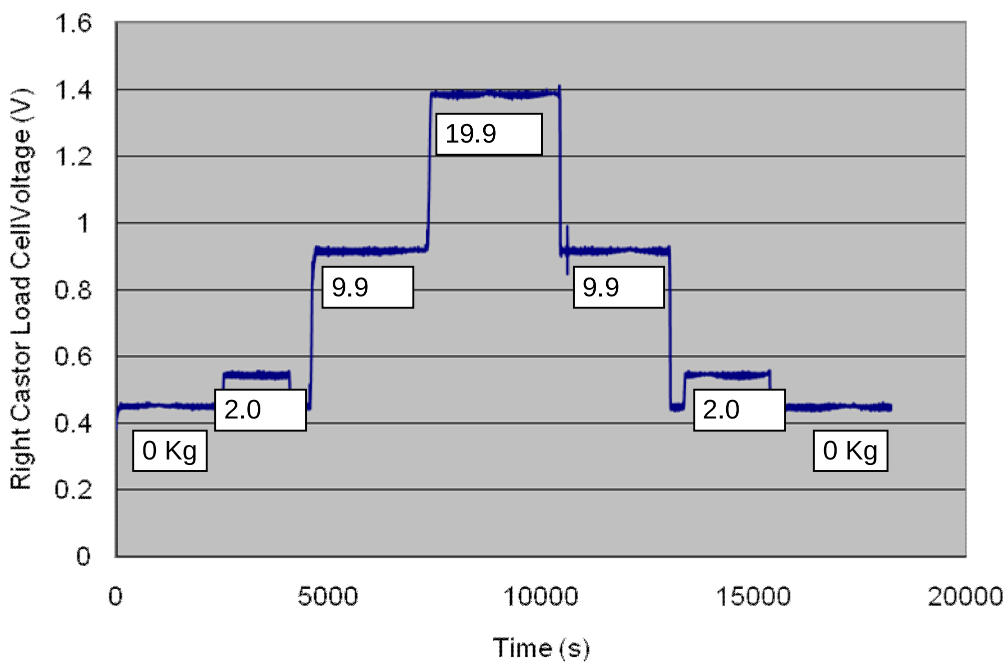


Figure 24: Calibration plot showing the known weights on the castor load cell.

Once data was collated this was transferred into Excel for analysis. This allowed the researcher to plot the information, using a scatter graph for the force place vs. force washer reading, to generate a calibration curve for the system. This calibration produced a hysteresis loop in order to demonstrate whether a linear relationship existed between the two force measurements and whether the loading-unloading behaviour of the system showed any hysteresis. Initial readings demonstrated extensive hysteresis, see (Figure 25).

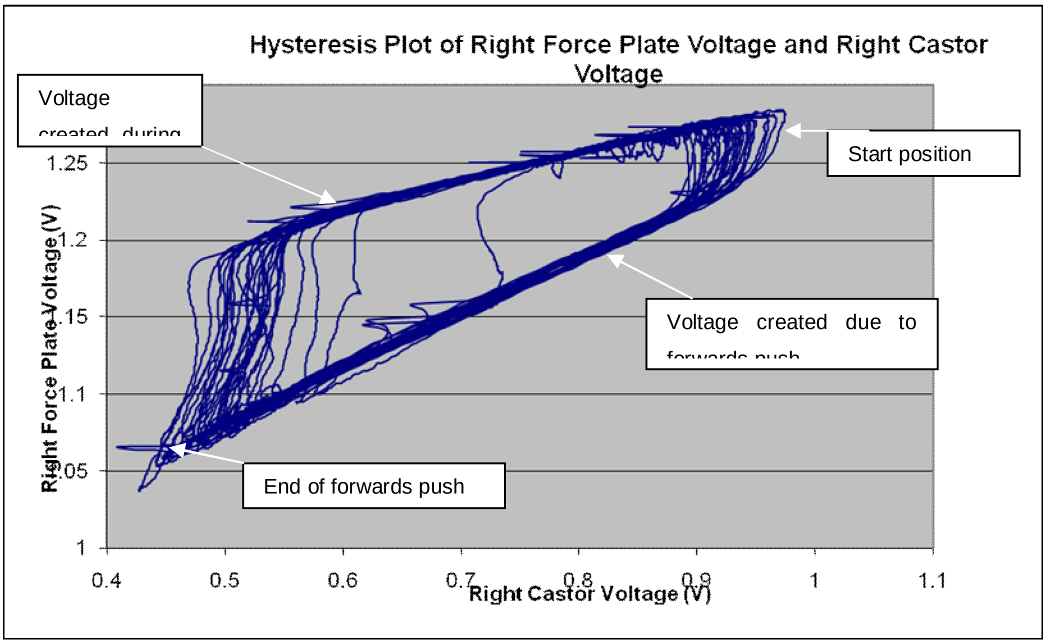


Figure 25 – Hysteresis Plot of right force plate and right castor voltage

Figure 25 shows the recorded propulsion cycle for the right force plate and the right castor. It has been labelled to show the direction of the propulsion cycle and is highlighting significant hysteresis present. It shows that the unloading of the castor force washer during a propulsion movement does not follow the same pattern as the forward loading movement.

In order to resolve this error there were a number of changes made to the castor construction including:

- use of rubber bungs and washers to reduce the friction
- use of a spring to reduce possible hysteresis in the rubber bung.
- repositioning of the force transducer
- repositioning of the bearing and washers
- adjustments to the spring tension through castor bolt adjustment

Considerations were made as to the direction of the forces generated in the castor. The trail and length of the castor fork influences the generation of friction forces as the stem passes through the bearing and therefore the forces transmitted to the force washer (Figure 26). Therefore, the castor fork was removed to identify the impact trail had on the reading. This demonstrated a linear relationship between the ground-castor forces and force washer forces and eliminated the hysteresis supporting the belief that friction between the stem and the bearing was caused by off axis loading of the bearing attributable to the trail of the castor fork.

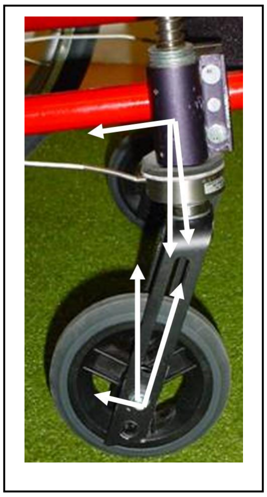


Figure 26 - Directional forces on the castor

Figure 27 is a repeated calibration graph with the castor fork removed showing a significant reduction in the hysteresis.

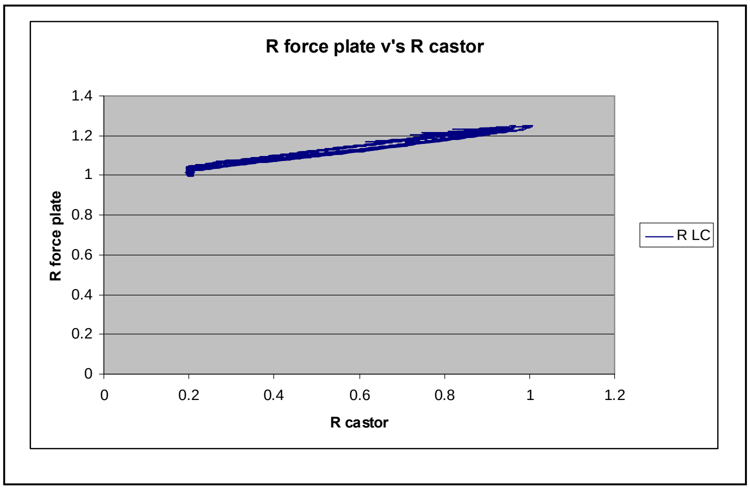


Figure 27 - Calibration curve without castor fork

Options to achieve the above results dynamically and with the castor fork in place included:

- Provision of a linear bearing within the castor housing.
- In house alterations to the castors to reduce the angle on the castor fork.

Due to the time constraints, the latter option was considered to be the most appropriate. Adjustments were made to the castor fork which statically generated excellent results but dynamically created problems with the propulsion of the wheelchair as the leverage generated by castor trail is important for easy of steering and maintaining straight-line directional stability(Figure 28).

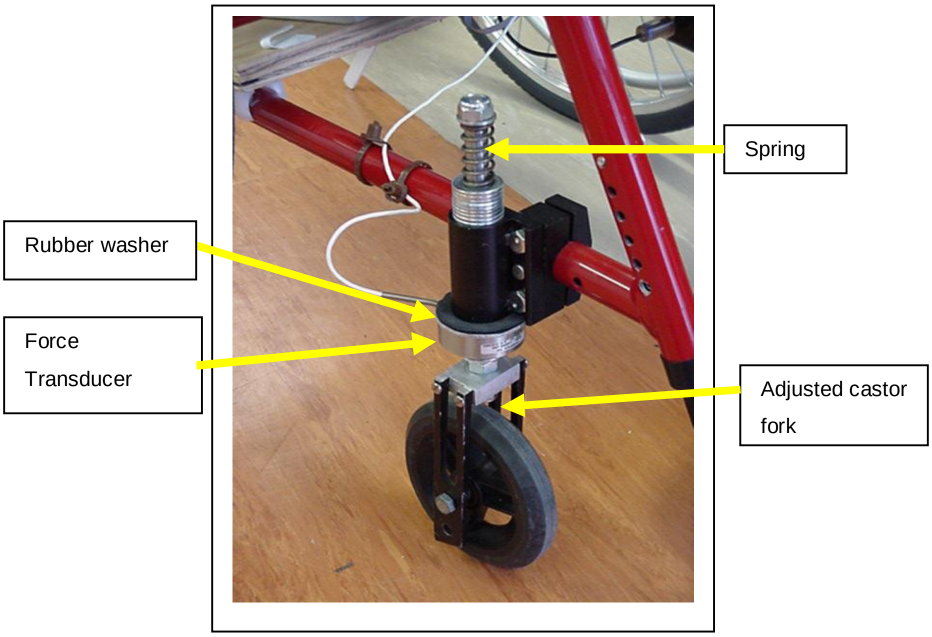


Figure 28 - 'In house' castor fork construction

As the construction of the castor fork did not have a trail there was extreme difficulty in turning the wheelchair. Therefore, a bearing was needed to accommodate the original castor fork, without introducing hysteresis.

The conventional engineering solution to accommodate off-axis loading in simple bearings is to use a ball bushing or linear bearing assembly. These come in simple axial forms which minimise friction along the axis of the rod but do not support friction-free rotation. A more complex bearing can be obtained that reduces friction in both axial and rotation (roller and linear bearing). However, it was not possible to obtain a roller and linear bearing to accommodate the castor stem without special order being placed and the time and cost implications were too great for this study. As long as measurements were to be obtained with the wheelchair travelling in a straight line the simple linear bearing was suitable. The manoeuvrability of the wheelchair was however compromised by the self-aligning feature of the linear bearing ï model - 0750-208-00 STD Precision Rexroth linear bushings (fitting 1/2" shafts) were used in an assembly. These are shown in Figure 29.

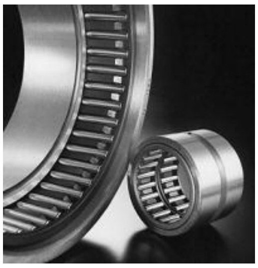


Figure 29 - Linear bearing

The final castor construction (Figure 30) used for the experiment is composed of:

(a) **Bolt** - The bolt was used to establish the correct amount of tension on the spring during calibration.
(b) **Spring** - The spring allowed the castor stem assembly to be tightened.
(c) **Castor housing** - . The castor housing enclosed the linear bearing as described above to reduce the amount of axle friction inside the housing from the stem.
(d) **Rubber washer** - The rubber washer restricted the movement of the force transducer around the castor stem during testing and ensured that the cable was not damaged.
(e) **Force transducer** - A force transducer was purchased (Interface Inc, model LW2050-250 capacity 20Ibf) and attached to the PDA to collect data of weight distribution.
(f) **Original fork castor** - The original castor fork was used but not as free to swivel due to the limitations of the linear bearing.

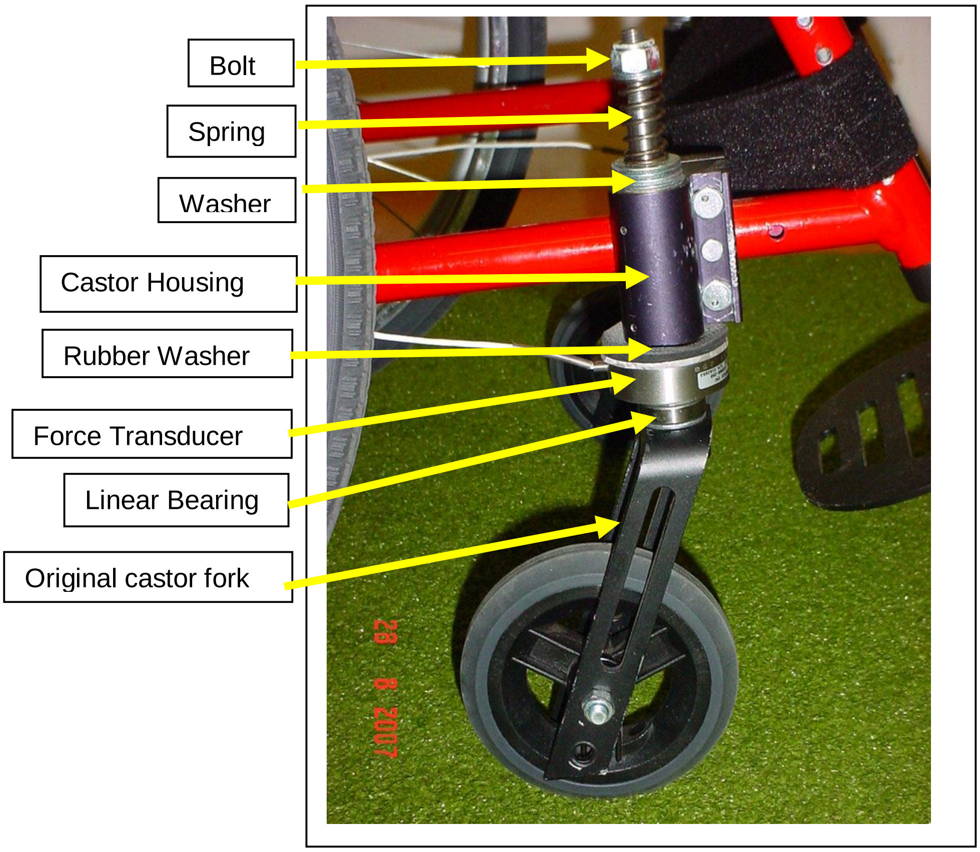
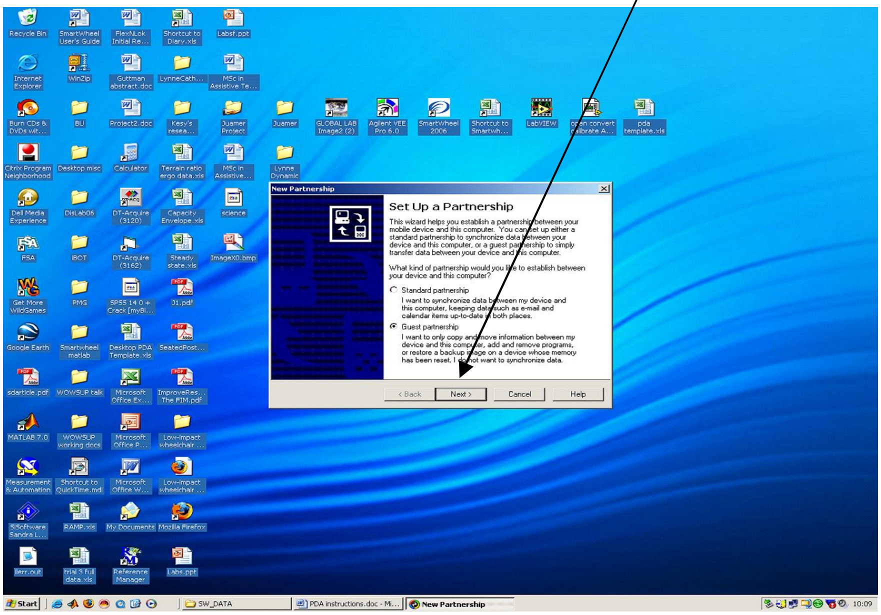


Figure 30 - Final castor configuration

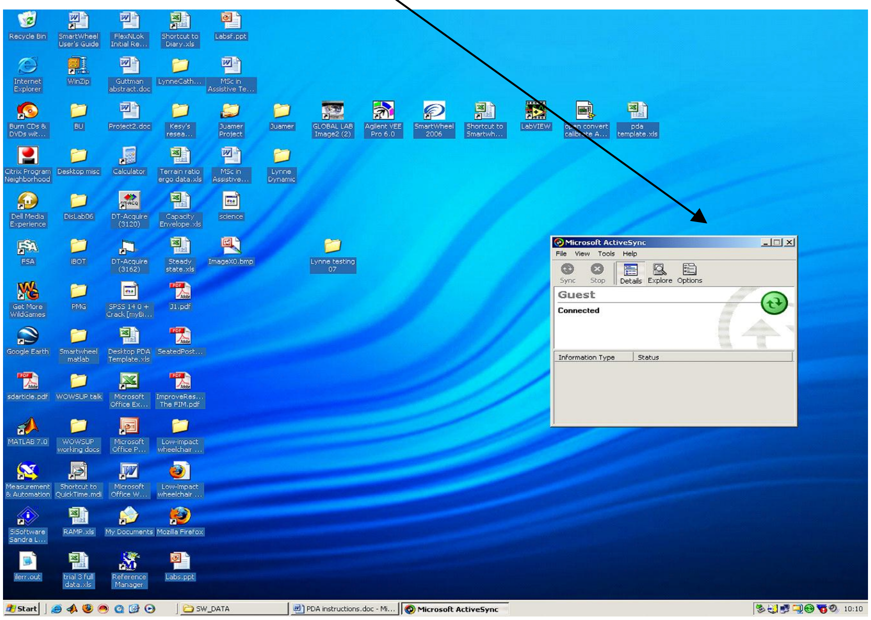
10 Appendix 3 - PDA Protocol

Plug into the DELL docking station ï make sure you exit the PDA programme (4ch 250Hz 02 2) before plugging into the docking station.

Set up Partnership ï Set up Guest Partnership - **Next**



Once guest partnership is set up it will display that this has been connected - **minimise**

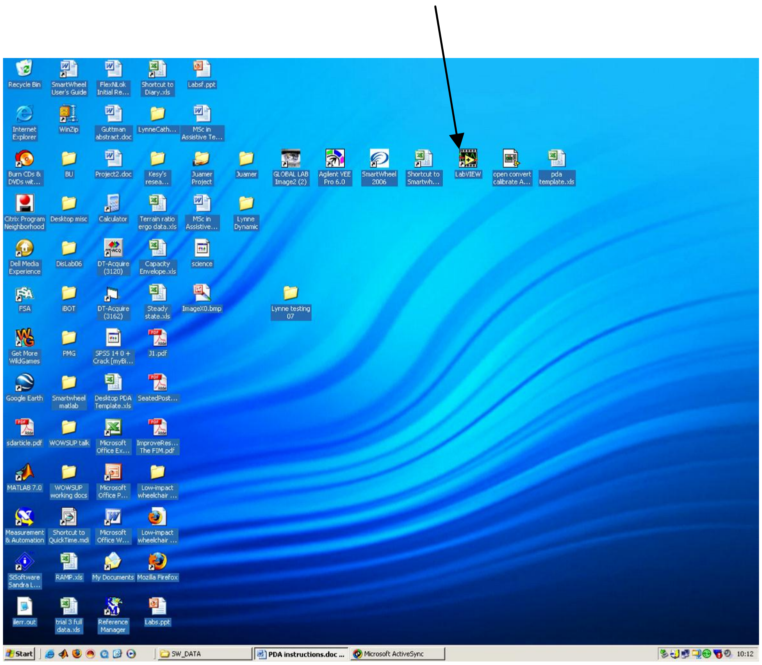


To obtain data file complete the following:

- My computer
- Mobile device
- My pocket pc
- SD card
- Martin

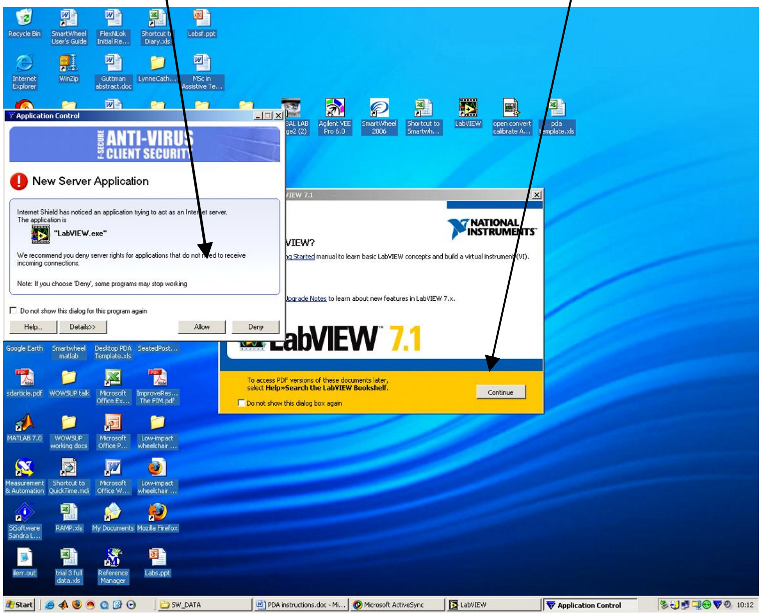
There files are not processed, they are all BIN files and need converting. Copy the ones you need and save them into your own file. Please delete all files once they are converted.

Go to Desktop and click on LabView



Allow Anti-Virus.

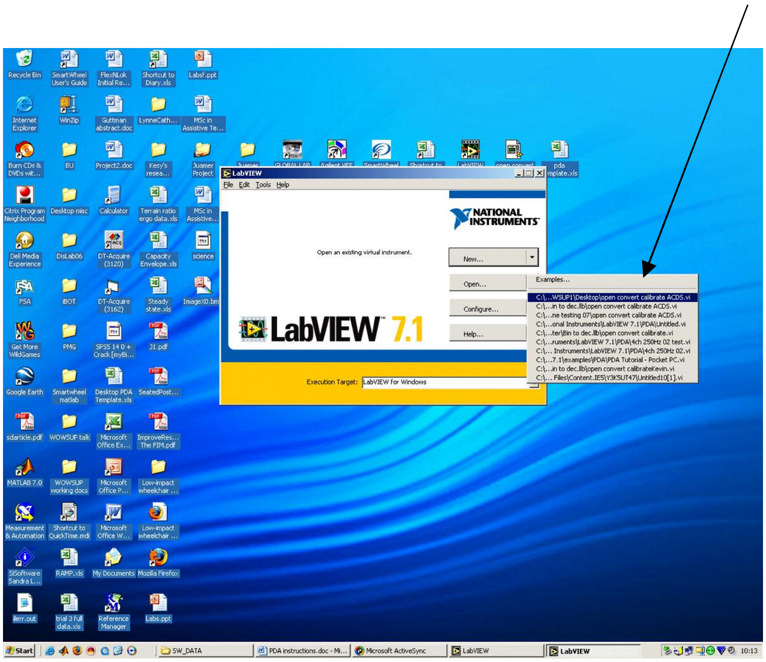
Click Continue on LabView 7.1



This will enter the LabView programme

Click on the Open drop down arrow

Click on C: \....WSUP\Desktop\open convert calibrate ACDS.vi

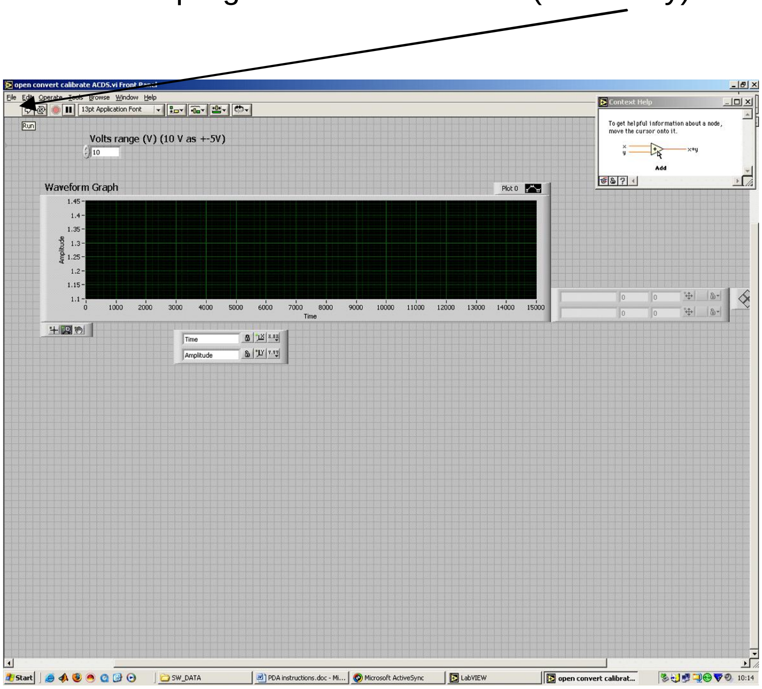


This is a short cut to the set up of programme.

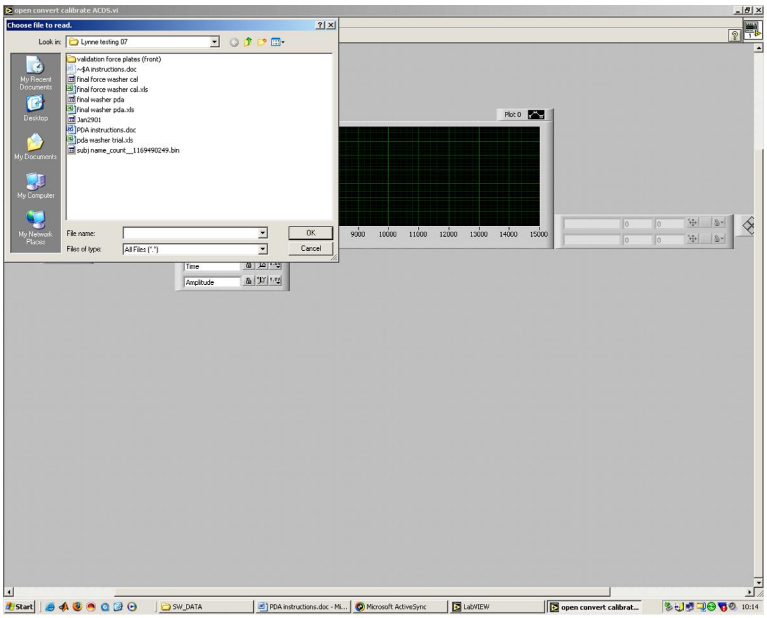
For the alternative route follow:

- Explorer
- Drive C
- Programme Files
- National Instruments
- LabView 7.1
- PDA
- Convertor
- Bin to deck

Once in the programme click on Run (arrow key)



Then identify which programme you want to convert and save it. It is worth calling it the same name but adding ćonôafter it so you know it is converted.

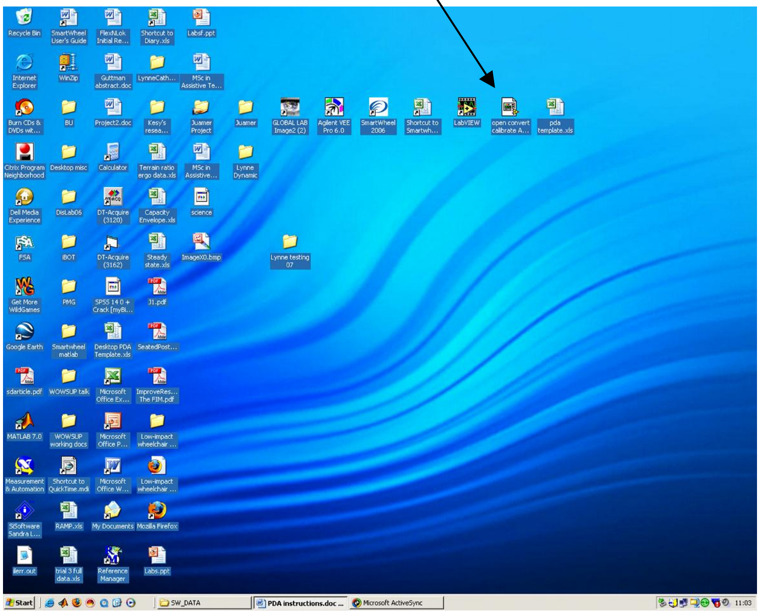


This will then provide you with the converted data in graph form.

These keys allow you to re-size/re-position the graph.

Short cut to PDA conversion

Click on open convert calibrate



Once the data is saved you can then transfer all the data into the template for analysis. This is held on the desktop named ôSwanky new graph templateô Double click on the icon and enable macros.

This will then ask you to IMPORT FILES ï select this and identify the required files to be imported. You can select up to 16 files. If you choose to transport less than 16 it may ask you to Debug ï select end and the graphs will fall on top of each other and will require sorting.

PLEASE DELETE ALL DELETED FILES ON PDA

11 Appendix 4 - Experimental Procedure

Participant into lab

Provide advice sheet and gain consent/sign form

Take required personal details

Set up in 17òGPV wheelchair

Position the axle in its rear most position

11.1 Smart Wheel Protocol

- Ensure the Smart Wheel is in the correct position ï positioned on the non dominant side.
- Run the wireless link for the Smartwheel on the laptop.
- Turn on Smart wheel switch with the brakes applied and wait for the double beep (important to have Smartwheel running before you launch software)
- Launch Smart wheel 2006
- Ensure that the Enable Trigger Input is checked
- The laptop should display the simple research view of live values ï close this down once open
- Go to the menu for Smart wheel 2006 (bottom right of screen on the toolbar) right click on this and select Smart wheel Session Wizard.
- Select Next
- Add Client
- And put in the information requested
- Choose the protocol (tile, carpet, etc)
- On the Trial Description page that comes up select ñauto-start the session wait for first push)
- Make sure the duration of the test is appropriate (e.g. 20s)
- Click Start Trial in the following order: Max Push, Lino, Astro, Ramp and Kerb.

11.2 Tachyon

Turn on the Tachyon box underneath the seat ï you will hear a dim buzzing noise

11.3 PDA

Turn on the PDA. Click on Start, then File Explorer and enter 4ch250Hz02-2 programme.
The time is set by highlighting the numbers under seconds per reading, clicking on the keyboard symbol (bottom right). Once the correct time is entered press the keyboard again to return to the main screen.

11.4 Accelerometer

(*white stick on rear of wheelchair*)
When completing any testing the accelerometer must be dropped to the ground at the same time that the Smart Wheel is started.

RECORDING and Protocol

11.5 Max push Protocol

MAX PUSH TEST – flat

- Ensure brakes are applied and the front castors are blocked with wooden blacks.
- Start up PDA and SW but wait 3 secs (neutral) prior to commencing actual pushing

PDA set to 15secs

- Ask Client to place their hands in their lap
- Instruct the Client to begin pushing *(Use the following script)*

Do not offer ANY encouragement to the client while they are pushing. After you stop the data collection, you may offer encouragement

"This test is designed to see the capacity of your push. When I tell you to 'GO' I want you to push your wheelchair as hard as you can with the brakes applied. The wheelchair should not move forwards. Please push for a count of 3 seconds with a rest of 2 seconds between each push. I will count and time you. When you have finished each push please place your hands on your knee. Do you have any questions?" *PAUSE* "Place your hands in your lap. GO."

- When subject starts the first push the trial and graphics should come up
- At the end of the run minimise the graphics view (you will only collect information for 20s)
- Select ñMake a Reportò
- Highlight ñmetric unitsò
- Tick the box for the file which you want to view (e.g. ñ04-05-2007 15-27 T Test (Tile protocol)
- Then click Build Word Report ï you can also view the visual data.
- The report will be generated. Save it as usual.
- The parameters definitions tab gives you a useful set of definitions and constraints for the parameters generated in the report.

Repeat x2 – following a break of 1 min between each attempt.

11.6 Lino Protocol

- Start up PDA and SW but wait 3 secs (neutral) prior to commencing actual pushing

PDA set to 15secs

- Ask Client to place their hands in their lap
- Instruct the Client to begin pushing *(Use the following script)*

Do not offer ANY encouragement to the client while they are pushing. After you stop the data collection, you may offer encouragement

"This test is designed to see how you push on a smooth floor. When I tell you to 'GO' I want you to push your wheelchair in a straight line. Push at a comfortable speed, as if you were pushing on a path. Keep pushing until I tell you to stop. Do you have any questions?" *PAUSE* "Place your hands in your lap. GO."

Smart Wheel turned on and accelerometer dropped to ground

- When subject stars the first push the trial and graphics should come up
- At the end of the run minimise the graphics view (you will only collect information for 20s)

Smart Wheel will stop recording following 20 secs and PDA will finish following 15secs – record time e.g 12:34:55

- Select ñMake a Reportò
- Highlight ñmetric unitsò
- Tick the box for the file which you want to view (e.g. ñ04-05-2007 15-27 T Test (Tile protocol)
- Then click Build Word Report ï you can also view the visual data.
- The report will be generated. Save it as usual.
- The parameters definitions tab gives you a useful set of definitions and constraints for the parameters generated in the report.

Repeat x2

11.7 Astro Protocol

- Start up PDA and SW but wait 3 secs (neutral) prior to commencing actual pushing

PDA set to 15secs

- Ask Client to place their hands in their lap
- Instruct the Client to begin pushing *(Use the following script)*

Do not offer ANY encouragement to the client while they are pushing. After you stop the data collection, you may offer encouragement

"This test is designed to see how you push across Astro turf. When I tell you to 'GO' I want you to push your wheelchair in a straight line. Push at a comfortable speed, as if you were pushing down a carpeted hall. Keep pushing until I tell you to stop. Do you have any questions?" *PAUSE* "Place your hands in your lap. GO."

Smart Wheel turned on and accelerometer dropped to ground

- When subject stars the first push the trial and graphics should come up
- At the end of the run minimise the graphics view (you will only collect information for 20s)

Smart Wheel will stop recording following 20 secs and PDA will finish following 15secs – record time e.g 12:34:55

- Select ñMake a Reportò
- Highlight ñmetric unitsò
- Tick the box for the file which you want to view (e.g. ñ04-05-2007 15-27 T Test (Tile protocol)
- Then click Build Word Report ï you can also view the visual data.
- The report will be generated. Save it as usual.
- The parameters definitions tab gives you a useful set of definitions and constraints for the parameters generated in the report.

11.8 Slope Protocol

- Start up PDA and SW but wait 3 secs (neutral) prior to commencing actual pushing

PDA set to 15secs.

- Ask Client to place their hands in their lap
- Instruct the Client to begin pushing *(Use the following script)*

Do not offer ANY encouragement to the client while they are pushing. After you stop the data collection, you may offer encouragement

"This test is designed to see how you push up a ramp. When I tell you to 'GO' I want you to push your wheelchair up this ramp. Push at a comfortable speed. You may rest if needed. Do you have any questions?" *PAUSE* "Place your hands in your lap. GO."

Smart Wheel turned on and accelerometer dropped to ground

- When subject stars the first push the trial and graphics should come up

Propel to top of the ramp – ensure assistant is behind

- At the end of the run minimise the graphics view (you will only collect information for 20s)

Smart Wheel will stop recording following 20 secs and PDA will finish following 15secs – record time e.g 12:34:55

- Select ñMake a Reportò
- Highlight ñmetric unitsò
- Tick the box for the file which you want to view (e.g. ñ04-05-2007 15-27 T Test (Tile protocol)
- Then click Build Word Report ï you can also view the visual data.
- The report will be generated. Save it as usual.
- The parameters definitions tab gives you a useful set of definitions and constraints for the parameters generated in the report.

Repeat x2

11.9 Kerb Protocol

- Start up PDA and SW but wait 3 secs (neutral) prior to commencing actual pushing

PDA set to 15secs

- Ask Client to place their hands in their lap
- Instruct the Client to begin pushing *(Use the following script)*

Do not offer ANY encouragement to the client while they are pushing. After you stop the data collection, you may offer encouragement

"This test is designed to see how you push up a kerb. When I tell you to 'GO' I want you to push your wheelchair up this kerb. If you are unsuccessful in this attempt then please return back to the starting point until further instruction. Do you have any questions?" *PAUSE* "Place your hands in your lap. GO."

Smart Wheel turned on and accelerometer dropped to ground

- When subject stars the first push the trial and graphics should come up
- At the end of the run minimise the graphics view (you will only collect information for 20s)

Smart Wheel will stop recording following 20 secs and PDA will finish following 15secs – record time e.g 12:34:55

- Select ñMake a Reportò
- Highlight ñmetric unitsò
- Tick the box for the file which you want to view (e.g. ñ04-05-2007 15-27 T Test (Tile protocol)
- Then click Build Word Report ï you can also view the visual data.
- The report will be generated. Save it as usual.
- The parameters definitions tab gives you a useful set of definitions and constraints for the parameters generated in the report.

Repeat x2

Download all data prior to re-testing

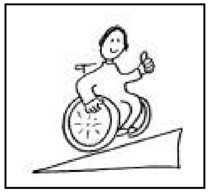
REPEAT ALL OF THE ABOVE WITH THE AXLE IN THE **MOST FORWARD** POSITION

12 Appendix 5 – Key Points for Clinicians

Calculating the risk of ‘Red Lining’

Functional Mobility
Ability to push on Lino, Astro, Ramp, Kerb Etc.

$$= \frac{\textbf{Performance} \text{ (Pushing on terrains)}}{\textbf{Capacity} \text{ (Max Push)}} \times 100$$



0%	Green	49%	50%	Amber	79%	80%	Red	100%

Increased risk of reaching capacity / ‘red lining’

<u>Effect of Functional Mobility Tasks</u>

Lino	Astro	Ramp	Kerb
			
• Increased velocity in stable set up • More likely to ‘red line’ in the stable set up	• Generating increased castor forces • ‘Red lining’ more likely to occur in the more stable set up at the start up phase of the run	• Increased push moment (Peak Mz) in the steady state • ‘Red lining' seen more frequently in the tippy wheelchair set up • Increased incidence of wheelchair users performing in the amber range of their capacity throughout the task	• Increased push moment (Peak Mz) in the more stable wheelchair set up on Astro and ramp • Requires skill rather than force

Effect of Rear Axle Position
Summary of Castor and Propulsion Forces

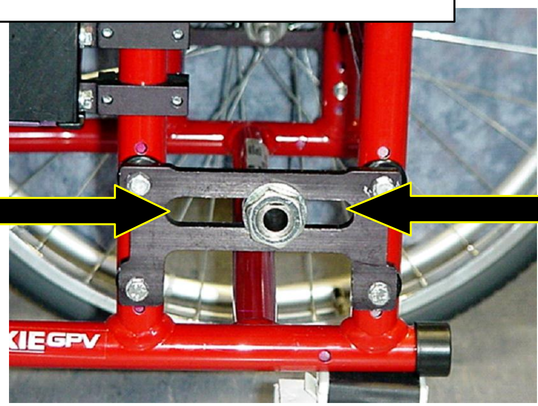
Propulsion Force

Stroke angle, v
and cadence remain the same on all terrains

Castor Force

⬆ Force on terrains imposing greater resistance

Influenced by trunk posture



Forward axle position 'tippy'

⬇ Castor forces

= Propulsion forces

Higher level injuries more likely to 'red line' on a ramp

Rearward axle position 'stable'

⬆ Increased castor forces

Increased velocity on lino

More likely to 'red line'

Improved smoothness of push

13 Appendix 6 - Glossary of Terms

ADL	Activities of Daily Living
ASIA	American Spinal Injuries Association Classification
Cadence	**Or push frequency.** This is defined as how many times per second, on average, the participant pushes on the SmartWheel rim during an entire trial.
CODA	A technology designed to capture 3D/4D activity
COM	Centre of Mass
DHSS	Department of Health and Social Security
Dynamic	Moving component
Functional Mobility Tasks	Everyday wheelchair propulsion activities
GH	Glenohumeral joint
Hand Contact	When the hand makes contact with the rim
Hand Release	When the hand releases all contact with the rim
ICF	International Classification of Functioning, Disability and Health
Kinematic	The study into how things move including forces, time and velocity.
Lightweight wheelchair	A wheelchair which offers adjustment and is made of lightweight materials

M/s	Metres per second
Max Castor Force	Highest castor force recorded during the propulsion cycle.
Max Push	Highest isometric push recording
MHRA	Medicines and Healthcare Products Regulatory Agency
Min Castor Force	Lowest castor force recorded during the propulsion cycle.
N/m	Newton metres
NHS	National Health Service
Optimal	Term widely used in the clinical setting to describe a wheelchair that has been set up to meet an individuals specific seating and performance needs
PC Ratio	This is a measure that has been designed to look at an individuals capacity to perform certain propulsion tasks. It determines whether an individual performs within their measured capacity or not.
Peak Average Force	The ratio between the peak force during a push, and the average force during a push. It provides an indication of how smoothly pushes are applied to the SmartWheel's pushrim.
Peak Mz	The peak propulsion moment that the participant applies to the SmartWheel during each push. This is the moment that turns the wheel (N/m)
Prime	Hands on the rim, push phase beginning also described by Kwarciak et al as initial contact (Kwarciak et al. 2009).

Push Phase	starts with a positive propulsion moment (Mz) and completed at hand release. Usually identified once contact is made with the push rim.
RAP	Rear Axle Position
Recovery Phase	starts immediately after hand release and is completed at hand contact
RPE	Rating Perceived Exertion
RR	Rolling Resistance
SCI	Spinal Cord Injury
SCRF	Stanmore Clinical Research Facility
SS	Last three pushes during a task
Stable	The wheelchair is usually set up with the rear axle in the most rearwards position meaning that it is less likely to tip backwards
Standard Wheelchair	A wheelchair which offers no adjustability
Start up	First Push
Static	Stable Component
Stroke Angle	This was defined as the angle travelled by the hand on the push rim from the point of contact to the point of release (in degrees). The average angle of the participant's push was recorded in degrees.
SW	SmartWheel
SWUG	SmartWheel Users Group

Tipping Angle	The angle away from the horizontal surface where a tipped wheelchair is critically balanced (Majaess et al 1993)
Tippy	The term ótippyô means óless stableô and is used throughout the thesis as it is a word commonly used clinically. It is a more neutral a statement than óstableô or óunstableô which have a connotation specifically in relation to safety. The axle is usually moved forwards in the chair allowing the user to perform advanced wheelchair skills with more ease.
Velocity	The average speed of the SmartWheel during each push. This can be used as an index of function. Average walking velocity is 1.4 m/s.
WCS	Wheelchair Service
WOWSUP	Workshop for Optimisation of Wheelchair Selection and User Performance