The Effect of Footway Crossfall Gradient on Wheelchair Accessibility

Thesis submitted to University College London for the degree of Doctorate of Philosophy

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I, Catherine Holloway, 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.

Abstract

This thesis investigates the effect of footway crossfall gradients (0 %, 2.5 % and 4 %) on wheelchair accessibility. This is done by instrumenting both a self-propelled and attendant-propelled wheelchair and asking a convenience sample of people to push the wheelchair in a straight line. Accessibility has been measured using the Capabilities Model. In particular the provided capabilities of the wheelchair users have been measured. These have been modelled as the interactions between the user and the wheelchair, specifically the amount of force it takes to start the wheelchair, the work needed to keep the wheelchair moving and the force needed to stop the wheelchair.

It is found that although the amount of work needed to traverse a footway remains constant regardless of crossfall gradient, a positive crossfall requires a second provided capability: the ability to apply different levels of force, and as a result work, to the upslope and downslope sides of the wheelchair. How people produce this difference of force is investigated. It is found that for self-propulsion, there are four strategies employed: the first is to reduce the force on the upslope side by pushing less hard, the second to increase the force on the downslope side by pushing harder, the third is to apply braking forces to the upslope wheel and fourthly to travel at a slower speed. These are either used independently or in combination. For the crossfall gradients tested it was found that attendants did not have to apply a negative (pulling) force to the upslope handle, and were able to combat the increased gradient by simply pushing harder on the downslope side.

The thesis concludes that current crossfall guidelines of 2.5% seem reasonable, and that inexperienced users may struggle when these guidelines are exceeded.

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1 Introduction

Being able to achieve goals is essential to an individual's quality of life. In the main, reaching a goal requires completion of one or more activities, each of which could be seen as being made up of a number of tasks. Each of these tasks must be possible for the individual to achieve in order for the activity, and thus the goal, to be accessible. Many of these activities take place away from a person's current location and thus it is often necessary to make a journey in order to be able to undertake an activity. For anyone to live a full and active life it is essential that they can participate in activities of daily living both in and outside of the home; for a wheelchair user this can present a major challenge. Their ability to leave their home to access services and participate in society therefore greatly impacts on their quality of life.

In the UK, wheelchairs are often funded by the National Health Service (NHS), who will also facilitate the adaptation of a wheelchair user's home to accommodate their needs and increase their ability to function within the home. However, the outside environment is less adaptable to the individual as it must be accessible to the majority of people. A basic skill of any wheelchair user[1], be they the attendant or occupant, is to be able to push a wheelchair along a footway. More often than not in developed countries such as the UK, footways will have a lateral slope (crossfall) to aid surface water drainage; of (it is recommended) not more than 2.5%[2]. However, there is little evidence for the current guidelines. It is important that such evidence be gathered.

Engineers and architects need to be confident that the built environment they design and construct is accessible to the majority of the public; to do this they follow accessibility guidelines. It is therefore essential that there is evidence to back up the guidelines. It is also important that the impact of not adhering to the guidelines is understood, as there will be occasions when it is impossible for them to be followed given physical constraints in the environment. Currently there is a split in opinion regarding the effect of crossfalls on wheelchair accessibility: the biomechanics research advises wheelchair users to avoid

[1] For this thesis when the term 'user' is used it refers to either the attendant or the occupant. When a specific user group is referenced the terms 'attendant' and 'occupant' will be used as required.
[2] Throughout this thesis percentage is used as a measure of footway crossfall gradient. Please see Appendix 1 for a table of conversions to degrees.

crossfalls where possible while civil engineering researchers are counselling a relaxation of the current U.S. guideline of 2%. This dichotomy is discussed in section 2.7.

This thesis will investigate the effect of crossfall gradient on wheelchair propulsion; both by the occupant and attendant. These two user groups make up approximately 85% of the 1.1 million wheelchair users in the UK, with attendant propulsion accounting for approximately 34% of users (Sapey *et al.* 2004). Attendant-propelled wheelchairs are provided to people who are predominantly unable to push themselves; these users are frequently elderly. Often the people who will push them (their carers) are spouses of the user or close friends. In these scenarios the carer would be of similar age to the occupant. Another large carer population are the children of elderly people; these attendants are generally over the age of 60. Therefore, a large proportion of carers will have their own health issues which can impact on their ability to push the wheelchair, which then directly affects the mobility of the wheelchair user (McIntyre & Atwal 2005).

The ways in which attendants and occupants push wheelchairs are fundamentally different. Self-propulsion of wheelchairs requires the user (the occupant) to release the handrim in order to move their arms back to the starting position of the push. Consequently, there is a period of time where the wheelchair is free to roll down the crossfall. This is not the case with attendant propulsion, where the person pushing has no need to release the handles. It is uncommon for a study to investigate both attendant and self-propulsion. In fact there have only been a handful of studies which have attempted to assess attendant-wheelchair propulsion (van der Woude *et al.* 1995; SUZUKI *et al.* 2004; Abel & Frank 1991). It is also unusual for a self-propulsion study to quantify the negative push-rim forces.

Whether or not a wheelchair user is able to push the wheelchair over a given terrain depends on their capabilities, and on the capabilities which result from the interaction between the user and the wheelchair. It also depends on the type of terrain being navigated and the interaction between the terrain and the wheelchair. In this thesis the approach which has been taken is to ignore the pure characteristics of the users (e.g. particular muscle strength) and those of the environment (e.g. surface hardness), and to focus on the interactions. This has been done within the framework of the Capability Model.

The Capability Model, which has been developed by Cepolina & Tyler (2004) is based on the philosophy of Capabilities and Functionings proposed by Amartya Sen. The Capability Model proposes that people have a certain set of capabilities (provided capabilities), which they can choose to use when they wish to do something. The activities the user chooses to do, and the environment in which they do them, have certain capabilities attached to them (required capabilities). For any given activity, the point at which the required capabilities exceed the provided capabilities signifies that the activity will not be achievable.

This thesis aims to develop a method for assessing both attendant and self-propulsion of wheelchairs in outdoor environments which may necessitate the application of negative forces. It is postulated in this thesis that by measuring the provided capabilities of the user one is able to make inferences about the accessibility of the footway. In this way mobility can be used as a measure of accessibility. The provided capabilities were reduced to the physical forces needed to successfully push a wheelchair in a straight line along a footway. It was hypothesised that when the footway was flat the user would need to provide 1) sufficient force to start the wheelchair, 2) sufficient work to keep the wheelchair moving over the required distance and 3) sufficient force to stop the wheelchair. When a crossfall was present these would need to be provided along with three additional capabilities: 1) a difference of force when starting, 2) a difference of work whilst going over the required distance and 3) a difference of force when stopping.

1.1 Aim & Composition of the thesis

The aim of this thesis is to measure the effect of footway crossfall gradient on wheelchair accessibility by measuring the provided capabilities of occupants and attendants. The thesis is divided into 9 chapters; the contents of these chapters are briefly described now.

Chapter 2 reviews the relevant background information including recent literature as well as developing the capabilities model.

Chapter 3 outlines the methods used to measure the provided capabilities of the self-propelled wheelchair system, when the wheelchair is being propelled over a three distinct crossfall gradients (0%, 2.5% and 4%).

Chapter 4 details the methods used to ascertain the provided capabilities of the attendant propelled wheelchair system, when the wheelchair is being pushed over a three distinct crossfall gradients (0%, 2.5% and 4%).

Chapter 5 reports the results of the experiments described in Chapter 3.

Chapter 6 reports the results of the experiments described in Chapter 4.

Chapter 7 discusses the results found in Chapter 5 and Chapter 6 in relation to the background information given in Chapter 2.

Chapter 8 details possible angles of further research given the results of the study.

Chapter 9 details the conclusions of this study.

There are also a number of appendices one of which I would like to bring to the reader's attention now. This is appendix 2, which contains a glossary of terms.

2 Background

This chapter begins by outlining the issues surrounding wheelchair mobility and accessibility (Section 2.1). In particular this section details the Capabilities Model (Section 2.1.1). The required capabilities imposed on a wheelchair user are then discussed (Section 2.2), followed by the provided capabilities of self-propelled users (Section 2.4) and then those of attendant propelled users (Section 2.5). The guidelines for footways are then briefly reviewed (Section 2.6) and finally some conclusions are drawn (Section 2.8).

2.1 Wheelchair mobility and accessibility

Mobility has been defined as “the ease of movement from place to place” (Tyler 2002; Tyler 2004). In this respect it can be seen as an interaction between how accessible the environment is (Accessibility) and the ability of someone to move in that environment (Movement) (Tyler 2002; Tyler 2004). Therefore, there are three things which interact to decide if someone can gain access to a place: the Person, the Environment and the Activity. The person chooses an activity they wish to do, which comprises of a number of tasks, each of which will occur in a certain environment.

People with mobility impairments often see the built environment as being composed of a series of barriers; things one encounters which hinder movement. The barriers that exist for wheelchair users are distinct from those which face people who are able to walk. Part of the reason for this lies in the fact the wheelchair must roll over the surface it is travelling on, whereas walking has a period of time when one or other of a person’s legs are raised from the ground. This difference accounts for the difficulty found by wheelchair users to traverse even small gaps and steps, which do not present a problem to most people when walking. The fact that a wheelchair rolls also means it has no immediate way to resist gravity when it is placed on a slope unless the brakes are applied. This is a particular problem for people who self-propel a wheelchair as this form of ambulation involves releasing the handrim for a period of time in order to allow the user to return their arms to the starting point of a push cycle; at this point, there is no force being applied to the wheelchair to counter the effect of the gravitational pull down the slope.

2.2 Measuring Accessibility: The Capabilities Model

Measuring accessibility is therefore complex as it requires knowledge of individuals' abilities and also an idea of how difficult different tasks are. In order to measure accessibility Cepolina and Tyler (2004) developed the Capabilities Model, which was further developed by Tyler (2006, 2009). In this model each task has a certain number of capabilities attached to it, required capabilities, which can change depending on the environment in which the task takes place. Each individual has a certain number of provided capabilities; cognitive, sensory and physical things they are able to do. For any given task, the point at which the required capabilities exceed the provided capabilities signifies that the activity will not be achievable.

The Capabilities Model for measuring accessibility is inspired by the welfare economics philosophy of Amartya Sen (Sen 1985, Sen 1993). Sen advocates that the quality of life of a person should be assessed in terms of 'his or her actual ability to achieve various valuable functionings as a part of living' (Sen 1993). Functionings are seen as the various things a person wishes to do or to be (Sen 1993). Since the publication of Commodities and Capabilities (Sen 1985), the capabilities approach has been adopted and discussed by researchers from a range of disciplines from philosophy to development studies (Anand *et al.* 2009). A core theme running through all of these studies is the distinction between the 'practical opportunities' available to people (their 'capabilities') and what they actually do (their 'functionings').

Measuring a person's capabilities is a more complex task than measuring the outcome of these capabilities and in the initial model a binary approach was taken to measuring the outcome of provided and required capabilities. The model enabled people to be identified as either being able to overcome a barrier, or not (Tyler, 2009). A barrier was defined in the model as the point at which the required capabilities exceed the provided capabilities and the task at hand became impossible for the person.

This binary version of the model was used by Cepolina and Tyler to develop a microscopic simulation of pedestrians moving about the built environment (Cepolina & N. Tyler 2004). The paper was illustrated with an example of three different pedestrians attempting to navigate through a gap and successfully demonstrated that accessibility can be modelled using the combined properties of the environment and barrier, (required capabilities) and the capabilities of the person (provided capabilities). Cepolina and Tyler concluded that the Capability Model as a concept provides a ‘good basis for the evaluation of accessibility’ of environmental barriers (Cepolina & Tyler 2004).

However, in this binary approach there is no indication of how difficult it is for the individual or how close they are to their maximum provided capabilities. A corollary to which is that failure can only be identified when it has occurred. An improvement on the model would be to enable prediction of failure based on data from when the participant was able to complete the task but was finding it more difficult.

Measuring the difficulty level of a task for an individual is complex (Tyler *et al.* 2007) approached this by measuring physiological responses to barriers. These physiological outputs are dependent on how someone completes the task. They give a continuous scale measure and therefore make it easier to compare the strategies taken by people. In this study, subjects’ heart rates were measured and an instantaneous and short term change in heart rate was observed which, in repeated tests appeared to be related to the participants’ responses to a change in the gradient of a footway. This work gave rise to the idea that such environment-person interactions might be measurable and that other, more appropriate, means of measurement should be investigated.

An added difficulty for researchers is that people may need to utilise different provided capabilities as task difficulty increases, or they might simply choose to use different provided capabilities. The difficulty for the researcher then, becomes distinguishing between these two cases (when they choose to change and when they must change). A starting point might be to assume that the change in experimental condition necessitated the change.

Traditionally transport engineers have assessed such things as trip length and number of journeys completed to assess how accessible a transport network is. These types of measurements constitute measures of functionings within Sen’s capability approach as they measure what people actually do and when they fail to do something rather than what they can do (capabilities).

The Capabilities Model represents a departure from this traditional approach to a more person-centred one. In particular this new model has at its centre the person who wishes to undertake an activity. However, this central role is one requiring the individual to take responsibility for the options open to them in accomplishing a task, and therefore take responsibility for at least some of their own capabilities. In this way it differs from the social model where society should be adjusted to the individual (and by extension all individuals) and from the medical model where the responsibility is removed from the individual.

2.2.1 Capabilities Model for Wheelchair Accessibility

In the case of this thesis the task in question is to propel a wheelchair in a straight line along a footway. As the task is fixed the model becomes a little simpler as we are dealing with only the interactions of the person and the environment. However, it is complicated by the addition of the wheelchair, and then further complicated by the addition of an attendant.

This system of interactions is shown in Figure 2-1, where the red arrows represent the required capabilities and the green arrows the provided capabilities. The black arrows indicated fixed capabilities within this thesis (as the task and environment are fixed).

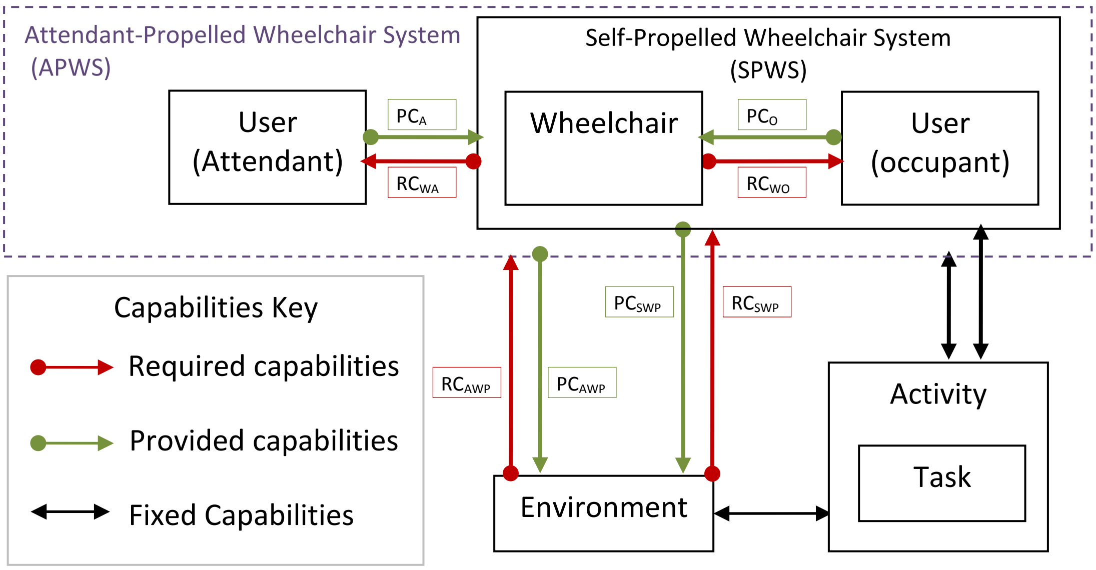


Figure 2-1: Interactions between the environment, wheelchair, users and activity using the Capability Model. Refer to text for details.

The Capabilities Model as shown in Figure 2-1 can be used to explore the problem of a wheelchair user wishing to do something which requires them to traverse a section of footway which has a crossfall. This involves investigating the various layers of provided and required capabilities that exist between the occupant, the attendant, the wheelchair and the environment.

In the simplest case when there is only one user of the wheelchair (the occupant), there is an interaction between the wheelchair and the occupant (see the top right corner of Figure 2-1). This can be represented as the set of provided capabilities the occupant has (PC_O) and the required capabilities needed to use a wheelchair by the occupant (RC_{wo}).

PC_O can include a whole range of abilities: cognitive, sensory and physical. In terms of wheelchair propulsion the main PC_O's are strength, fitness and technique. These exist independently of the wheelchair. The Wheelchair has a number of Required Capabilities, which exist without for the need for an interaction with the user or the environment, and consist in general of the design and set-up of the wheelchair. The interaction of PC_O and RC_{wo} produce a net set of provided capabilities, PC_{SWP}, which then interact with the Environment's required capabilities (RC_{SWP}). The environment itself and all of the barriers it contains represent the required capabilities, which must be overcome in order for the task to be achieved.

Therefore, improving the accessibility of a task can occur by increasing the occupants provided capabilities (e.g. a training course to improve their fitness, strength or technique). Alternatively, it can be achieved by decreasing the required capabilities of the wheelchair (e.g. altering its set-up to make it easier to push). Lastly accessibility can be improved by reducing the required capabilities of the task (e.g. making the surface smoother or reducing the crossfall gradient).

When the occupant of a wheelchair is unable to propel themselves and are pushed by an attendant, it is then the attendant's provided capabilities which determine if a task can be completed, while the occupant remains part of the required capabilities in a purely mechanical way (as they are a part of the mass which needs to be moved). This is modelled in Figure 2-1. The output of the provided capabilities of the attendant (PC_A) and the required capabilities of the SPWS (RC_{WA}), which must be pushed by the attendant, result in a net provided capabilities of the Attendant-Propulsion Wheelchair System (APWS). These must then be greater than the required capabilities of the environment (RC_{AWP}), in order for a task to be accessible.

2.2.2 The importance of Footways to Accessibility

In order to access any part of a town or city in the developed world, it is necessary to traverse a footpath. This is made clear by the European Conference of Ministers of Transport:

"Almost all journeys start and finish by walking or wheeling. No matter how accessible transport itself may be, if the walking [or wheeling] environment contains barriers to movement than the usability of transport services is largely negated"

(European Conference of Ministers of Transport 1999)

Footways form an integral part of the built environment worldwide. Many countries have introduced standards to ensure pavements do their job; to provide a safe and effective surface for people to use in order to access the buildings and services. Initially footways would have been used simply to walk along and to ensure they remained free from surface water, which not only causes problems regarding safety (people can slip in wet or icy conditions) but also structural issues such actions as the 'freeze thaw' action of water. This causes microscopic cracks to become bigger over time through the expansion of freezing surface water. However, in more recent times the needs of those who have some kind of mobility impairment need to be considered. This group consists of those who would have traditionally been thought of as being 'disabled' and as such need a form of assistive technology to aid them in traversing the pavements (such as wheelchair users and those who require a walking stick or crutches to keep their balance), to those who are impaired through their choice of shoes or amount of luggage they have decided to carry, or the child they need to push.

The consequence of this is that, accessibility forms a large part of whether or not an individual is socially excluded and social exclusion has been shown to be linked to the accessibility of public transport (Church *et al.* 2000; Hine & Mitchell 2001). Tyler highlighted that there is a particular need for pedestrians in London using the bus system to be able to traverse 400m of footway. This is the straight-line distance used by London bus companies to a bus route (Tyler 1999). This means that on average a pedestrian will only be 400m away from a bus route at any point in London. It was found by Barham and colleagues that only 15% of wheelchair users were capable of propelling 360m to a bus stop without needing to rest, this figure increased to 40 % when the distance was halved to 180m (Barham *et al.* 1994). The recommended distance for wheelchair users to travel without a rest is 150m (Department for Transport 2005). The footways in this study were not graded in terms of slope or surface type.

2.3 Required Capabilities: Wheelchairs & the Environment

As has just been discussed, the required capabilities are made up of the interaction between the wheelchair and the environment. The type of wheelchair and how it is set-up along with the topology of the terrain can all increase the required capabilities, making the wheelchair more difficult is to push. These factors will now be explored, starting with the types of wheelchairs (2.2.1). Then defining and discussing factors which affect the wheelchairs rolling resistance (2.2.2). Finally some of the common methodologies for measuring rolling resistance are briefly discussed (2.2.3).

2.3.1 Measuring Rolling Resistance

In sections 2.2.2 the theoretical framework for investigating the rolling resistance of wheelchairs over varying topographies has been investigated. This framework assumed that the rolling resistance does not change with velocity. This is not strictly true as both the deformation properties of tyres (Kauzlarich & Thacker 1985)and also the rolling mechanics of a wheelchair (Chua *et al.* 2010; Hoffman *et al.* 2003) can change with velocity. Furthermore, while theoretically there should be an increase in rolling resistance with crossfall angle, there is no published data which empirically proves this. For this reason two simple tests were performed to investigate the effect of velocity and crossfall angle on the rolling resistance of a wheelchair.

There are a number of methods one could use to measure the rolling resistance of a wheelchair. One of the most popular is to use what is called a ‘drag test’ where a wheelchair is attached to a treadmill via a rope (de Groot *et al.* 2006; van der Woude *et al.* 1986). Attached to one end of the rope is a 1-dimensional force transducer. Therefore at any given velocity the force transducer is measuring the force that the wheelchair is resisting. The test has many advantages: the speed of the wheelchair can be controlled accurately through the settings on the treadmill, the wheelchair will travel in a relatively straight trajectory and the rolling surface is smooth. However, it has one drawback as it can only measure the interaction between the wheelchair and the treadmill surface (M. D. Hoffman *et al.* 2003). There can also be differences between different treadmills and different set-ups of the tests as were highlighted by (de Groot *et al.* 2006)**.**

A second methodology is the 'push-bar technique', where a force transducer attached to an attendant propelled wheelchair is used to measure the rolling resistance (van der Woude *et al.* 2003). In this type of test someone pushes the wheelchair while attempting to limit the amount of vertical force applied to the wheelchair**.** This test methodology was adapted from Glaser & Collins who were interested in calculating power output from wheelchair users by measuring the average force needed to push a wheelchair and occupant from behind and multiplying it by the velocity of the wheelchair (Glaser & Collins 1981).

Using the handle-bar push technique it has been found that the force required to keep a wheelchair moving over everyday indoor terrains varies from approximately 10 N (on tile) to 30 N (on high-pile carpet) (van der Woude *et al.* 2003).

Using a variation[3] of the push-handle technique the rolling resistance of concrete pavers was tested both when the surface was flat and when there was a 2.5% and 4% crossfall. It was found that the rolling resistance of the surface was approximately 24 N, and the effect of crossfall gradient was to increase the rolling resistance by approximately 3.4 N per percentage increase in crossfall gradient.

2.3.2 Wheelchairs: An Introduction

Wheelchairs provide an alternative mode of mobility for those who find it difficult, or impossible, to walk. A wheelchair consists of a chassis which houses a seating unit for the user to sit. It should be noted the seating unit also provides the stable base from which self-propelled users can push the wheelchair. A wheelchair has a number of wheels. In most cases wheelchairs have two larger rear wheels and two smaller front casters which are free to rotate. Wheelchairs can be divided into three main types: self-propelled, attendant-propelled and electric. This division is based on the method of force application to the wheelchair in order to make it move.

[3] The variation involved using an electric scooter to pull the wheelchair rather than a person push the wheelchair. A full methodology as well as the results of this experiment is given in Appendix 3:.

The focus of this thesis is on manually powered wheelchairs. It will focus on two types of wheelchair, the standard wheelchair prescribed by the National Health Service (NHS) for attendant propulsion and, secondly, the rigid-framed wheelchair often prescribed for self-propelling users with high activity levels. The latter group of users is often referred to as Active Users.

Figure 2-2 illustrates the percentage of 'clients'[4] who use different types of wheelchairs depending on their mobility and postural needs. 55% of clients generally use a wheelchair as their main form of mobility and Active Users are a subset of this. The attendant-propelled

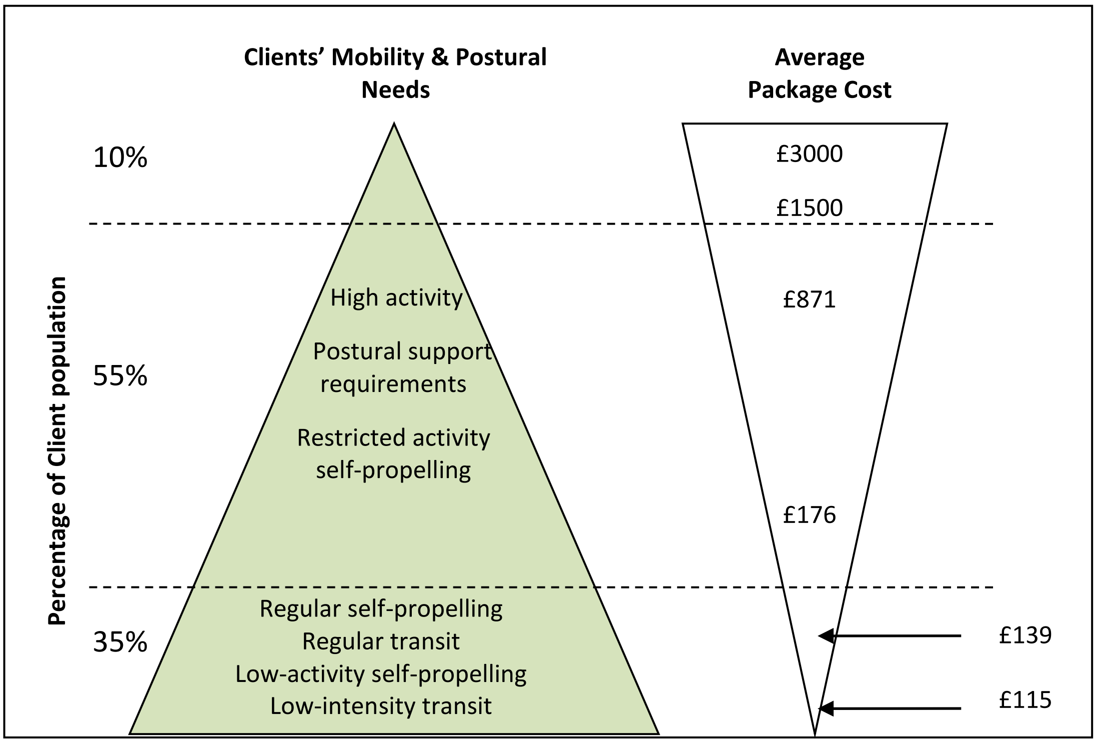


Figure 2-2: The considerations in mobility and postural management provision. Adapted from Le Grand 2008

users and self-propelled users who do not use their wheelchair daily, account for approximately 35% of users according to Le Grand (2008) (see Figure 2-2). This figure for the number of people who are pushed by an attendant concurs with the figure (34%) reported in a recent survey in North West England into the social implications to the increased use of wheelchairs in society (Sapey *et al.* 2004).

[4] The term client refers to people who are assessed for a wheelchair by the National Health Service

It can be seen from Figure 2-2 that there is a substantial difference in the average package cost[5] of a wheelchair provided mainly for attendant propulsion (£115-£139) and one for an active user (£871). These reflect the cost of design and manufacture of the different wheelchairs, which will be discussed briefly now.

The 'standard' wheelchairs prescribed by the NHS for attendant-propulsion are the 8L and 9L (the equivalent wheelchair in the U.S. are the K1001 and K1000). Both wheelchairs weigh approximately 18kg. The 8L and 9L are both wheelchairs with a folding, steel tubular design. As can be seen in Figure 2-3, which shows an 8L, these wheelchairs have canvas seats and backrests, with removable footplates and arm rests. They are able to turn thanks to 'shopping trolley' style casters at the front and have rear facing handles to allow for attendant propulsion. The only difference between them is their rear wheel size; the 8L has larger rear wheels which allow the occupant to self-propel, while the 9L has small rear wheels.

The larger rear wheel of the 8L theoretically makes it easier to push (something which will be examined in more detail in section 2.2.2). However, it also makes it wider and so more difficult to manoeuvre through doorways and more awkward to put into the boot of a car.

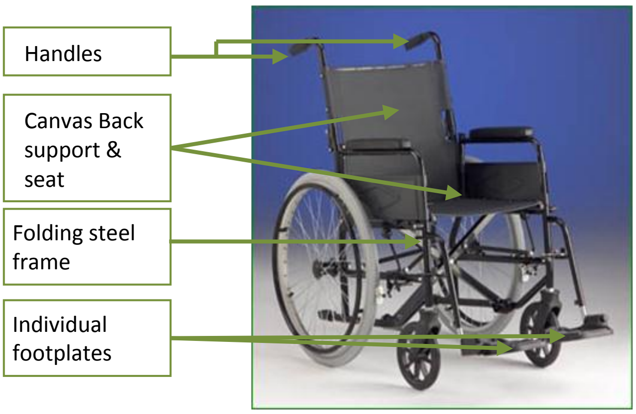


Figure 2-3: Standard issue wheelchair for attendant propulsion

Importantly though the large rear wheels afford the user the opportunity to propel themselves, even if they only have the ability to move small distances, and perhaps only indoors. For this reason it is often prescribed for both frequent and occasional attendant propelled use as well as for those who will self-propel regularly. Neither the 8L nor the 9L

[5] The 'package cost' would include items such as cushions, specialist seating systems which help maintain upper-body posture and also costs incurred in personalising the wheelchair.

are dissimilar in design from the first folding frame wheelchair designed by Herbet & Jennings in 1933 (Sawatzky n.d.; Kamenetz 1969) .

In stark contrast the current design of wheelchair for active users has a number of adjustable features, which allows for a greater degree of customisation to the individual (Michael L. Boninger *et al.* 2000). One of the major factors in allowing such a high degree of customisation has been the rigid-frame, which has been the platform for what can be seen as a revolution in wheelchair design (Lucas H.V. van der Woude *et al.* 2006). This design change eliminates internal energy losses, which occur in the folding frame design due to flexing of the frame (Lucas H.V. van der Woude *et al.* 2006; Michael L. Boninger *et al.* 2000; L. H. V. van der Woude *et al.* 2001). A key element in the fixed-frame design is that the seat height and position can be altered as can the rear axle position. The significance of these alterations to wheelchair rolling mechanics is that they change the distribution of the weight, which can in turn make the wheelchair easier to push on any given surface. This will be addressed in section 2.2.2.1. They also affect how much of the handrim it is possible for the occupant to grasp, which in turn can change the kinetics and kinematics of the push cycle (Michael L. Boninger *et al.* 2000; Cowan *et al.* 2009).

The change in frame design has been accompanied by an improvement in material technologies (DiGiovine *et al.* 2006; van der Woude *et al.* 2006). At the forefront of design are materials such as titanium and carbon fibre (DiGiovine *et al.* 2006; van der Woude *et al.* 2006). Carbon fibre is seen as offering huge potential in future wheelchair design, although this is yet to be realised (DiGiovine *et al.* 2006). It allows for the possibility to add rigidity to a wheelchair in one direction (prevent the frame flexing laterally) and flexibility in another direction (provide a suspension system vertically) depending on the orientation of the fibres (DiGiovine *et al.* 2006) . Titanium offers a high strength-to-weight ratio and is capable of absorbing shocks and vibrations. However, due to raw material costs and increased costs in machining, wheelchairs made of titanium are more expensive than the more standard aluminium or steel wheelchairs (DiGiovine *et al.* 2006).

It is for this reason (cost) that it is not common for the NHS to prescribe titanium wheelchairs for 'active' wheelchair users. The standard is instead an aluminium rigid frame wheelchair, commonly referred to as a 'lightweight chair', it is equivalent to a 'K0005' in the U.S.. This wheelchair offers a compromise between the ultra-light Titanium wheelchair and the standard folding steel frame wheelchair. Aluminium is cheaper and easier to machine than Titanium, while having a higher strength-to-weight ratio than steel (DiGiovine *et al.* 2006).

Despite the lack of titanium, the aluminium wheelchair can still be considered 'a task specific, versatile functional device' (van der Woude *et al.* 2006) due to the large number of adjustments it allows for. These adjustments improve the wheelchair in three ways: firstly they allow for a set-up which optimises the push of the occupant (Kotajarvi *et al.* 2004), secondly they can be used to relieve pressure on areas prone to pressure sores by being able to adjust the seat and finally they can be adjusted to reduce the rolling resistance of the wheelchair. The concept of rolling resistance (the force a user must overcome to make the wheelchair move or remain moving) is important in the context of this thesis and so will be looked at now in greater depth.

2.3.3 Wheelchair Motion & Rolling Resistance

A wheelchair will remain at rest, or continue to move at a constant velocity, unless a force acts upon it according to Newton's First Law of Motion. How efficiently a wheelchair is pushed depends on the users' capabilities along with: the weight of the wheelchair system (occupant and wheelchair), the way in which this weight is distributed and the friction between the wheelchair and the rear wheels and casters (Brubaker 1986). These last three things can be considered independently of the user (Brubaker 1986).

The weight of the wheelchair system is directly proportional to the frictional force created between the wheelchair and the rolling surface. The frictional force (f) that needs to be overcome can be calculated from the normal reaction force (R) and the coefficient of friction (μ) with the following formula: $\mathrm{f} = \mu . \mathrm{R}$.

The normal reaction force is equal and opposite to the weight of the object to be moved. The value of μ is higher when an object is being accelerated from rest (referred to as the coefficient of static friction) compared to when it is already moving (called the coefficient of dynamic friction). Therefore the heavier the wheelchair system and the more friction there is, the greater the force necessary to move the wheelchair. In the case of wheelchairs the frictional forces that occur between the wheelchair and the ground are termed rolling resistance; as the wheelchair rolls over the ground.

In the case of a flat surface the forces acting on a wheelchair at rest can be seen in Figure 2-4A. As the wheelchair is at rest the only force acting on the wheelchair is the combined weight (W) of the wheelchair and the user, indicated by the red vertical arrow. This produces reaction forces at the contact points with the ground (R_c at the casters and R_w at the wheels). When a force is applied to the wheelchair (this could be applied by the occupant, F_w, to the handrim or to the handles by an attendant, F_h, or both) frictional forces (F_c at the casters and F_w at the wheels) will oppose the motion of the wheelchair (Figure 2-4B).

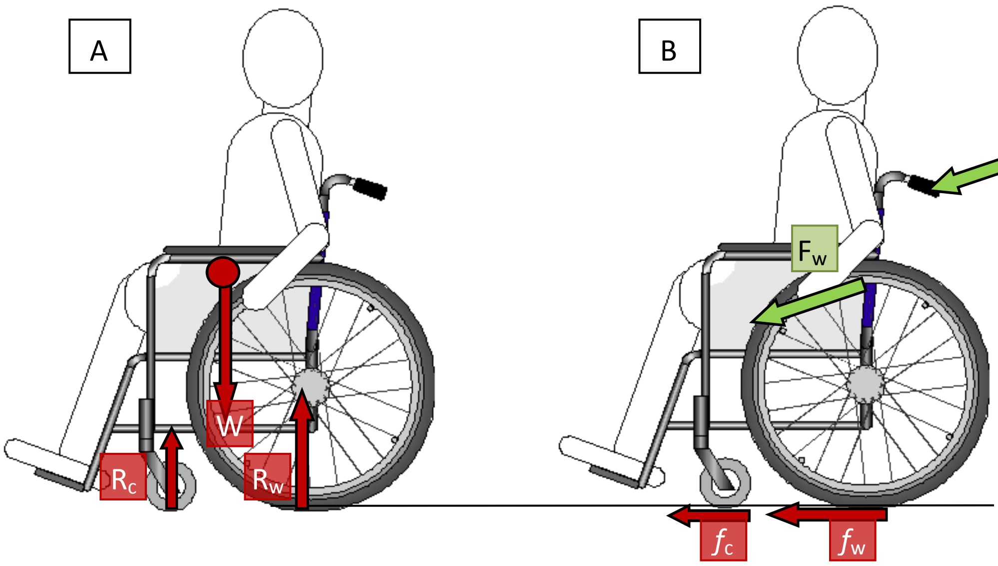


Figure 2-4: Forces acting on a stationary wheelchair when no forces are being applied by a user (A) and when force(s) are being applied by a user (B). See text for description.

2.3.3.1 The effect of wheel and caster properties on rolling resistance

The magnitude of each resistive force occurring at the wheel or caster occurs due to deformation of the tyre surface, as well as deformation to the surface. When travelling on concrete footways the deformation of the surface can be considered negligible, whereas when travelling on a surface such as sand it becomes significant. The magnitude of the frictional force also depends on the radius of the wheel, the material of the tyre and the weight travelling through the caster or tyre. The rolling resistance of a wheel is inversely proportional to the radius of the wheel (McLaurin & Brubaker 1991; van der Woude *et al.* 2001)(Brubaker 1986). The rolling resistance of a wheelchair can be expressed as:

$$F_{RR} = \mu_w R_w + \mu_c R_c$$

Equation 1: Where F_{RR} is the rolling resistance force, μ_w represents the combined coefficient of friction of the wheels, R_W is the combined normal reaction force at the wheels, μ_c represents the combined coefficient of friction of the casters and R_c is the combined normal reaction force at the wheels.

The fact that with all other factors being equal, smaller radius wheels have a higher coefficient of friction means that, theoretically, wheelchairs should be designed and set-up so as to minimise the amount of force being transferred through the casters. In lightweight wheelchairs this is possible by adjusting the vertical height of the seat and adjusting the rear axle position. A 2.5 cm adjustment of the rear axle position (from 10 cm to 7.5 cm) has been shown theoretically by Tomlinson to reduce the rolling resistance by 10% (Tomlinson 2000a)[6]. Experimentally, this has been backed up by Boninger *et al.* who found that moving the rear axle forwards so as to unload the caster resulted in 'better propulsion biomechanics' (Michael L. Boninger *et al.* 2000). By 'better' the authors are referring to taking fewer, longer and less abrupt pushes, thus reducing the likelihood of injury to the upper limb (Michael L. Boninger *et al.* 2000). Pushrim biomechanics and their relationship to upper limb injury will be addressed in greater detail in Section 2.3.2.

Attendant-propelled wheelchairs can have much smaller rear wheels than self-propelled wheelchairs; with a radius of approximately 25cm compared to 60-65 cm. These smaller rear wheels are usually solid rather than pneumatic, something which also increases their rolling resistance compared to pneumatic tyres (Sawatzky *et al.* 2004). They also tend to be thicker, which increases the contact profile with the ground (in much the same way as a flat pneumatic tyre), which further increases rolling resistance. Furthermore, there is no adjustability of rear axle position on the attendant-propelled wheelchair and the wheelchair itself weighs more (by approximately 8 kg). Therefore, theoretically, it can be concluded attendant-propelled wheelchairs have a higher rolling resistance than their lightweight counterparts. What is more difficult to conclude is whether the larger rear wheels of an 8L offer a higher resistance than the smaller 9L wheels, given that the 8L can be solid (but generally not of the same material as the 9L) or pneumatic. There has been no published study that could be found which quantified this difference.

[6]By reducing the rolling resistance in this way, the wheelchair would also be made less stable making it easier to flip the casters and perform a wheelie but also to fall backwards, which can result in injury and even death of users(Brubaker 1986; Calder & Kirby 1990; Michael L. Boninger et al. 2000).

The orientation of the front caster also affects its resistive force; when it is inline it decreases and when at 90 degrees to the direction of travel it provides maximum resistance to propulsion (van der Woude *et al.* 2001). The caster is an interesting component of the wheelchair as it is passive and is free to rotate about the axis in its housing. It is this component that allows the wheelchair to swivel and turn. It is also responsible for allowing the wheelchair to turn down-slope when on a crossfall.

Finally, the rear-wheel camber angle can be changed, this would technically change the ground reaction force due to a change in contact area between the wheelchair and the ground. A study could not be found to quantify this effect, which is probably due to the fact the effect is minimal. Changing the camber angle though does have additional possible benefits for wheelchair users traversing a footway with a crossfall, as there is an increase in the distance between the turning centre of the wheelchair and the contact point of the wheelchair and the ground. Therefore, a greater camber should help resist the downward turning moment on a crossfall (this turning moment is described in detail in section 2.3.3.3). This has been backed up in experimental work by Trudel et al who found changes to the camber angle improved both the manoeuvrability and stability on slopes (Trudel *et al.* 1995; Langner & Sanders 2008). Increasing the camber angle will also increase the wheelbase, which may make manoeuvring in tight spaces more difficult (Langner & Sanders 2008).

The effect of slopes on wheelchairs will now be investigated along with design and set-up features that either aid or hinder a wheelchair's motion along a sloped terrain.

2.3.3.2 The effect of sloped terrain on wheelchair motion

When a wheelchair is on a longitudinal slope, in other words going up or down a hill, the user must overcome not only the rolling resistance which is present during flat terrain propulsion, but also the additional force of gravity (Richter *et al.* 2007a). The magnitude of this additional force ($F_{incline}$) is dependent on the weight (W) of the wheelchair system and the slope of the terrain, which is expressed mathematically as:

$$F_{incline} = W.\sin\theta$$

Equation 2: The force required to traverse an incline ($F_{incline}$), where W is the weight of the wheelchair system and Θ is the angle of incline.

This additional force is shown in Figure 2-5 as a blue arrow. This additional force is added to the rolling resistance found on the flat condition (Equation 3).

$$F_{Tot} = RR + W.\sin\theta$$

Equation 3: Total force (F_{tot}) needed when travelling up a slope, where W is the weight of the wheelchair system and Θ is the angle of incline and RR is the rolling resistance as defined in equation 1.

In the case of travelling up a hill extra force must be applied to the handrim or handles to prevent the wheelchair rolling back down the hill. However, when travelling downslope, this force due to gravity can be beneficial for wheelchair users, as the force due to gravity is acting down the slope. In this case it is akin to an additional propulsive force and it accelerates the wheelchair. When thought of in this way the energy used to climb a hill can be 'harvested' during the descent (van der Woude *et al.* 2001). The rolling resistance is thus reduced (see Equation 4).

$$F_{tot} = RR - W.\sin\theta$$

Equation 4: Total force (F_{tot}) needed when travelling down a slope, where W is the weight of the wheelchair system and Θ is the angle of incline and RR is the rolling resistance as defined in equation 1.

However, the additional force aiding acceleration when travelling downslope will need to be overcome if it becomes necessary, either for safety or comfort, to slow or stop the wheelchair.

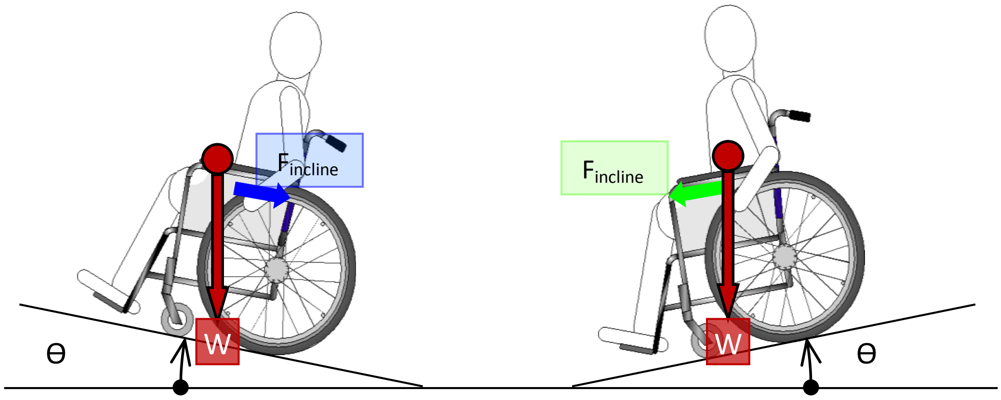


Figure 2-5: Forces acting on a wheelchair going up (left) and (down) a slope. W is the weight of the wheelchair system and Θ is the angle of incline.

2.3.3.3 The effect of crossfalls on wheelchair motion

Crossfalls are commonplace on UK footways; their presence aids the drainage of surface water off the footway and into the roadside gutter. However, the same force (gravity) also attempts to pull the wheelchair downslope (see Figure 2-6). Although the rear wheels will resist this lateral force, the casters do not (Richter *et al.* 2007a). Therefore, a turning moment is created, with the force (Fc on Figure 2-6) acting at the centre of mass of the wheelchair system perpendicular to the line OC shown on Figure 2-6. The line OC is the line connecting the centre of mass and the midpoint between the two rear wheels (Richter *et al.* 2007a). The resulting moment can be quantified by Equation 6.

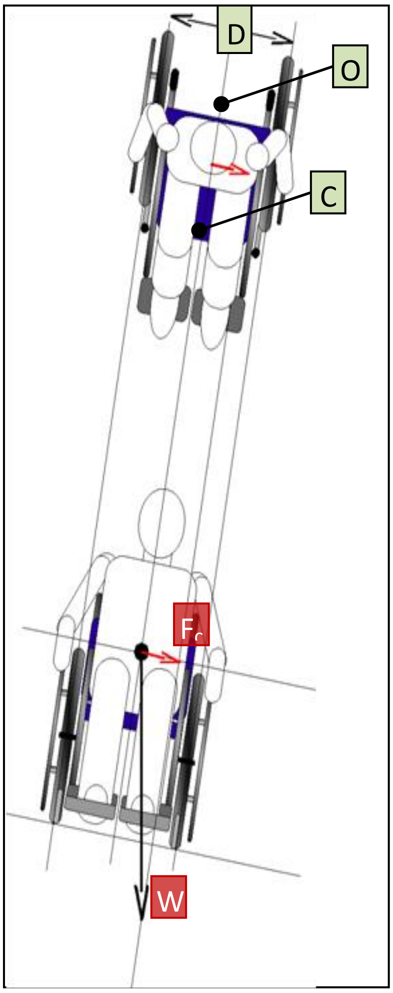


Figure 2-6: Illustration of a wheelchair on a crossfall. Please see text for full description of terms.

$$M_{cfall} = W.\sin\theta.d$$

Equation 6: Equation for the static moment (M_{cfall}), which acts on a wheelchair when it is at rest on a surface with a crossfall. Where W is the weight of the wheelchair system, θ is the angle of the crossfall and d is the distance between the contact points of the two rear wheels and the ground.

To overcome this moment the user must apply an equal and opposite moment to M_{cfall}. This is achieved by applying a difference of force between the upslope and downslope sides of a wheelchair. This difference of force can be achieved by applying a braking force to the upslope side of the wheelchair, or an increase in force on the downslope side, or a combination of both. Whatever the details of the force application, a force difference must exist between the upslope and downslope sides in order to counter the force created by the crossfall and for the wheelchair to continue in a straight line. The magnitude of this force is calculated with the same formula as in the case of the longitudinal force (see Equation 2). However in this case the angle (Ө) is the angle of crossfall gradient, not the angle of the longitudinal slope.

One thing that must be highlighted is the fundamental difference in force application of self-propelled wheelchairs compared to attendant-propelled wheelchairs. Self-propulsion requires the user to release the wheels during what is called the recovery phase of the stroke. During which time the wheelchair is free to roll down slope. This can of course be prevented by constantly applying a braking force to the upslope wheel. However, this force is only aiding forward motion of the wheelchair in an indirect way and is not in fact actively producing work (i.e. the force does not cause the wheelchair to move in the direction of desired travel.

2.3.4 Work and Energy

Work is an important physical parameter as it shows the result of the force being applied and the distance achieved by the application of the force. In order to do work in a mechanical sense one must apply a force to an object and move it. Therefore work is calculated using Equation 7, and its unit is the Joule (J), where 1J =1Nm.

$$\text{Work} = \text{Force x Displacement}$$

Equation 7: Equation for work done on an object

In the case of a wheelchair user applying a force to the wheelchair in order to prevent it rolling down a crossfall; the amount of work they achieve on the wheelchair by simply preventing the wheelchair turning downslope is zero, as they are preventing movement. This is of course if they are successful. However, this does not mean that the person preventing the motion of the wheelchair has not expended energy.

The details of how humans produce work is beyond the scope of this thesis, as it would require a focus on measuring the displacement of joint segments (e.g. the upper and lower arm) relative to each other. The reason this is necessary is that the relative displacement of each joint segment would need to be known along with the forces applied to both segments in order to calculate the work. This is not done in this thesis. However, it is important to realise that when a force applied to a wheelchair is not aiding the intended direction of propulsion, it is still having a physiological cost on the person applying the force.

Work has recently been used as an objective measure of surface firmness (Chesney & Axelson 1996). Using a Smart Wheel (see Section 2.4.1 for a full description of the SmartWheel) the amount of work needed to cross different surfaces was measured and averaged to get ‘work per meter’. They also tested a number of different crossfall conditions using this measure, the results of which are reported in section 2.8.

2.4 Provided capabilities: Self-Propulsion

The SmartWheel (SW)[7] is a commercially available tool for measuring handrim forces and moments, as well as wheel speed and stroke angle. It attaches to the axle receiver of a wheelchair in place of a standard wheel[8].

The self-propulsion of wheelchairs is a strenuous task compared to walking, with a gross mechanical efficiency of approximately 10% (Groot *et al.* 2002). Therefore it is necessary to maximise the provided capabilities of the Self-Propelled Wheelchair Systems (SPWS). In order to do this the force applied to a wheelchair needs to be measured, this is made possible with a SmartWheel, which will be described in Section 2.3.1). This will be followed by a description of the key kinematic parameters for everyday environmental barriers faced by wheelchair users 2.3.2. Finally injuries linked to manual wheelchair propulsion will be discussed 2.3.2.3.

2.4.1 SmartWheel

One of the driving forces for developing the SmartWheel (SW) had been to collect research data that was “clinically meaningful, useful and practical” (Cowan *et al.* 2008). In addition it allowed non-traditional research centres to participate such as hospitals and clinics. There are a growing number of SW users, many of whom contribute to the SmartWheel Users Group (SWUG). UCL is one such institute. This group was set-up to help pool data from all studies involving the SW. To enable similar data to be collected worldwide a SW Clinical Protocol was developed. The first publication to result from this data pooling was published last year (Cowan *et al.* 2008). It highlighted 4 parameters from the 21 that the SW can automatically generate to be the most ‘clinically important and relevant’ (Cowan *et al.* 2008). These are speed, average peak resultant force, push frequency and stroke length and all are calculated during ‘Steady- State’. Steady-state is defined as occurring after push 3 has occurred in the SmartWheel Clinical Protocol. A detailed description of each parameter is given in Table 1.

[7] The SmartWheel is developed by Three Rivers Holdings, LLC.
[8] The technical specification of the SmartWheel is discussed in greater detail In section 3.6.3

Table 2-1: Description of SmartWheel parameters adapted from the SmartWheel User Guide 2008 (Cowan *et al.* 2008).

Parameter	SmartWheel User Guide Description
Push Frequency [1/s]	This is how many times per second, on average, the Subject pushes on the SmartWheel.
Average Peak Resultant Force [N]	The resultant force is calculated mathematically by combining the 3 orthogonal force components measured by the SW. A peak value is produced with each push and these are averaged for each trial.
Average Speed [m/s]	This is the average speed of the SmartWheel during steady state (the time after the first 3 pushes).
stroke length [m]	The length of push stroke
Average Peak Tangential Force [N]	The average peak tangential force. This is the force component which is tangential to the handrim

These parameters are excellent for describing straight line propulsion; however they are unable to produce a complete picture of how users accomplish more complex tasks such as completing a figure of 8-track, something which is acknowledged by Cowan *et al.* (Cowan *et al.* 2008).

2.4.2 Wheelchair Pushing

Wheelchair self-propulsion consists of a cyclical push pattern, which necessitates a period of time when the hand is not in contact with the handrim. This results in what is termed a push phase followed by a recovery phase (Sanderson & Sommer III 1985; Kwarciak *et al.* 2009). Kwarciak *et al.* point out this is equivalent to the swing and stance phases of gait (Kwarciak *et al.* 2009). This research expands the definition of the push phase into three distinct parts: the initial contact, the propulsive phase and the release of the handrim. It gives a clear and precise definition of where each phase starts and ends see (Kwarciak *et al.* 2009). Their push phase definition remains consistent with previous work; occurring when a positive driving moment (Mz[9]) is applied to the handrim.

Generally researchers simply report on the push phase, with each using a different method of finding the start and end of the push. Some have used 0 Nm cut-off on the M_z signal (S. de Groot *et al.* 2002; Hurd *et al.* 2008), some have used a positive resultant force (F_{res}) (Sabick *et al.* 2004; Richter *et al.* 2007b) while others have identified the start and end of a

[9] Throughout this thesis when referring to self-propulsion it is assumed that the rear axle of the wheelchair constitutes the z-axis and that a positive moment causes forward motion. Therefore,Mz is the positive moment about the rear axle.

push by searching from within the push until F_{res}=0 or dF_{res}/dt =0 (Richter *et al.* 2007a). This has made it difficult to compare between studies, something which is highlighted by (Kwarciak *et al.* 2009).

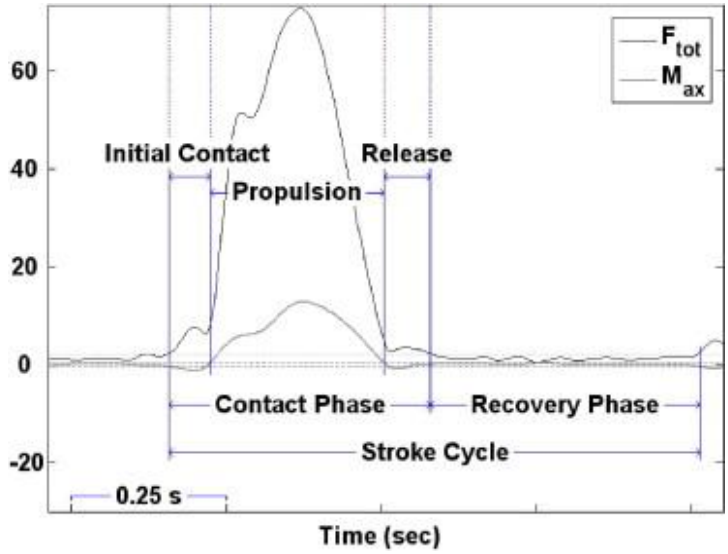


Figure 2-7: Graph of Ftot (equivalent to Fres) and M_{ax} (equivalent to Mz) showing the start and end points of the new stroke cycle definition, taken from (Kwarciak *et al.* 2009).

In conclusion, there are a variety of methods used by people to identify the start and end phase of a push phase used by researchers, but the definition of using a 0 Nm or 0N cut-off seems fairly well established and logically consistent with what one would expect a push moment or force to be. The new cut-offs proposed by Kwarciak are probably more valid provided the noise is normally distributed on the signals (Kwarciak *et al.* 2009). The reason for mentioning this in so much detail is that when propelling on a crossfall it can become even more difficult to define the start and end of a push due to the user leaving their hand in contact with the handrim during what would normally be termed the recovery phase. This will be addressed in Section 1.3.

2.4.2.1 Start-up

During straight line wheelchair propulsion, initially the wheelchair accelerates and then achieves a period of what is termed 'steady state' propulsion and then decelerates to a stop.

The start-up phase is normally reported as being composed of the first three pushes and the steady-state phase occurs from push four (Three Rivers 2008; A. M Koontz *et al.* 2005). When research is carried out on an ergometer or treadmill it is quite easy to collect sufficient data to identify these two states. However, when data is collected in the field this becomes more difficult and in some cases it becomes necessary to only use the fourth push in the steady-state phase calculations (Cowan *et al.* 2008).

When an ergometer is used, steady state is normally defined as beginning once a self-selected or target velocity has been achieved, and ending arbitrarily 20 seconds later (Kwarciak *et al.* 2009; Michael L. Boninger *et al.* 2000). The end point can vary; being measured in metres (Aissaoui *et al.* 2002), or measured for longer periods of time e.g. (Rick N. Robertson *et al.* 1996). This study used a 30 seconds time window and then analysed 5 consecutive pushes in the middle of this data (Robertson *et al.* 1996). There have been studies to suggest that there is a high degree of variation between pushes and that the only way to counter the variation is to increase the number of pushes taken for analysis (Rodriguez *et al.* 2004). Rodriquez *et al.* showed that increasing the number of pushes taken for analysis from 3 to 30, from a single subject on a treadmill, reduced the coefficient of variation for the peak force by 60.9 % (Rodriguez *et al.* 2004).

Start-up studies remain fewer in number than those analysing steady state parameters. However, the identification of the start of this phase is easier; it is then only a question of when this phase shifts to steady state. One way of identifying the end of the start-up is to run separate repeated-measures analysis of variance tests on each of the biomechanical measures of interest and then use pairwise comparisons to see which strokes appear statistically similar, and which appear different. This was the approach taken by Koontz *et al.*, who found that of the 7 propulsive moments for which biomechanical measures had been calculated pushes 1-4 were statistically different from 5-7 (Koontz *et al.* 2005). Furthermore, the first three pushes accounted for the majority of the start-up phase with push 4 being a transition from start-up to steady state (Koontz *et al.* 2005). Therefore, they defined steady state as starting from the forth push and it is frequently the case that the first three pushes are ignored in studies wishing to concentrate on the steady state phase.

2.4.2.2 Laboratory versus 'real world'

De Groot *et al.* investigated the impact of task complexity on the mechanical efficiency of wheelchair propulsion over nine weeks of training (de Groot *et al.* 2008). The three levels of task complexity were dependant on the type of wheelchair propulsion. The least complex was propelling on an ergometer, a treadmill was considered intermediately complex and the most complex was propelling around a track. The wheelchair was only capable of measuring forces on one side and so the total power was estimated by multiplying by 2 for the ergometer and treadmill, and by combining alternative runs for the track condition.

The study conducted by Koontz *et al.* (1995) was one of the first to investigate the effect of outdoor surfaces and was also one of the first to utilise the SmartWheel. It identified that people adapt the input force magnitude and frequency in response to different rolling resistances, caused by the different surfaces. Then, if necessary, they reduce their speed. At least this could be one interpretation of the fact the participants in Koontz et al's study increased the total number of pushes, and the peak tangential force when travelling over moderately high rolling resistance surfaces. However, on the ramp and grass, people also went more slowly. It could be concluded that these two surfaces offered too much resistance for the user to counter in order to reach their desired velocity.

As was pointed out by Koontz et al higher forces have been linked to increasing upper limb injuries in regular manual wheelchair users. It would appear that people have a certain velocity in mind and try to reach that velocity on all surfaces. Start-up forces, regardless of surface, caused higher peak forces and moments than the steady-state phase.

2.4.2.3 Injuries linked to pushing parameters

Upper limb injury affects a high proportion of manual wheelchair users (MWU). The two main sites of injury are the wrist, which is prone to carpel tunnel syndrome (Gellman *et al.* 1988; Boninger *et al.* 1999; Boninger et al 2004a; Aljure *et al.* 1985; J. Yang *et al.* 2009) and the shoulder, which is prone to impingement syndrome and rotator cuff tears along with aseptic necrosis at the head of the humerus (Bayley *et al.* 1987). The pain caused by upper limb injury to manual wheelchair users can be debilitating and has been likened to the equivalent of a higher level injury in spinal cord injured (SCI) patients by Sie and colleagues (I. H. Sie *et al.* 1992; Boninger et al 2004b). Severe pain has also been linked to a lower quality of life score of SCI patients in the four years post injury, in fact it was the only complication factor found in 99 patient histories that correlated to a lower quality of life score (Lundqvist *et al.* 1991; Boninger *et al.* 1999).

Carpel Tunnel Syndrome (CTS) is a common injury found in non-wheelchair users and wheelchair users alike. It is caused by repeated movements and has been found in ergonomic studies to be linked to tasks requiring high range of motion of the wrist, which are done repeatedly. The incidence of CTS among MWU's has been reported as being between 40% and 86% of the MWU population. It has been found in all studies concentrating on SCI patients to increase with the number of years post injury (Gellman *et al.* 1988; I. H. Sie *et al.* 1992; Aljure *et al.* 1985). As manual wheelchair propulsion is a repetitive movement requiring considerable wrist range of motion, the high incidence of CTS in MWUs is not surprising. These ergonomic findings led Boninger *et al.* to investigate the effect of wrist range of motion on ulnar and radial nerve function Boninger *et al.* 2004. They tested nerve function by doing a Nerve Conduction Study (NCS) which involved stimulating the ulnar and radial nerves at the wrist and then recording the time it took the signal to travel to the second and fifth digits of the hand muscle which are controlled by these nerves, the amplitude of the measured signal was also recorded (Boninger *et al.* 2004). The results of the NCS were correlated to kinematic data, which had been collected with a SmartWheel. Based on the ergonomic literature, the researchers had expected that an increase in the range of motion in the wrist would result in reduced amplitudes of ulnar and radial nerve stimulus, representing poorer nerve health. However, it was found that significantly higher ulnar and radial nerve amplitudes were found with increases in wrist range of motion (Boninger *et al.* 2004). On further inspection it was found that push cycles that used a larger range of motion in the wrist also resulted in fewer pushes to attain the same velocity, along with reduced peak forces (Boninger *et al.* 2004). Based on these findings the authors recommended a 'longer smooth stroke' as the best method to preserve healthy nerve function when propelling a manual wheelchair (Boninger *et al.* 2004). This would suggest that, as a minimum, peak forces, rate of loading of force and push frequency should be recorded when analysing kinematics if one wishes to make inferences about the propensity for possible wrist pain resulting from nerve injury caused by wheelchair use.

2.5 Provided Capabilities: Attendant Propulsion

The provided capabilities of attendant wheelchair propulsion consist of how well an attendant can impart force to the wheelchair handles. The provided capabilities are thus the output of the interaction between the wheelchair and the attendant (see Figure 2-1). What follows is a review of literature pertaining to the measurement of the provided capabilities needed to successfully push and pull a wheelchair.

2.5.1 Pushing and Pulling

When pushing the force is directed away from the person or body applying the force and when pulling the force is directed towards the origin of the force (Hoozemans *et al.* 1998).

A study conducted in 1991 (Abel & Frank 1991) stated there was no published data which supported the design of wheelchair handles that were being used; this is still true today. In fact the interaction between the attendant and wheelchair has barely been looked at all. The handle height, its stability and its orientation are all factors that affect how much force is required to accomplish a particular task. In general a higher handle height is preferred for pulling than pushing, with no recommendations for handles above shoulder height (Table 2-2).

Within the Ergonomic and Occupational Biomechanics literature there has been a considerable amount of research done into the effect of handle height on the ability of people to apply maximal force. Todd, in his review of the trends occurring in research focussed on pushing and pulling, comments that despite the large number of studies there seems to be no consensus on what is 'optimal' (Todd 2005). This could be because each

Table 2-1: Table of handle height findings and recommendations taken from Todd 1995

Authors, date	Findings	Recommended handle height
Martin and Chaffin (1972)		50 – 90 cm above the floor
Ayoub and McDaniel (1974)	Increase H increases Fmax	Pushing = 91 – 114 cm Pulling = 94 – 115 cm
Warwick *et al.* (1980)	Push: increase H increases Fmax Pull: reduce H increases Fmax	
Chaffin *et al.* (1983)	Decrease H increases Fmax	Between shoulder and hip
Gagnon *et al.* (1992)	Reduce H increases Fmax	
Lee *et al.* (1992)	Push: Increase H increases Fmax Pull: reduce H increases Fmax	
De Looze *et al.* (1995)		
Kumar (1995)	Fmax greatest at middle H	
Fothergill *et al.* (1999)	Pull: reduce H increases Fmax	

study has a slightly different focus, for example some are measuring compression forces on the lumbar spine, while others measure the peak handle forces.

2.5.2 Pushing and injury

Much of the work done in investigating pushing and pulling has varied in methodology and in general has not received as much attention as lifting, which is seen as the primary risk factor in lower back injuries within the work place (Jansen *et al.* 2002). A particular concern for wheelchair propulsion is that if the carer is unable to push the wheelchair and achieve functional mobility for the wheelchair user, then they will simply not use the wheelchair and this will in effect leave the user without a form of mobility. There is no direct proof of this in the literature. However, when talking to wheelchair engineers and occupational therapists as well as nurses and other healthcare staff it seems they believe there is a risk of this occurring.

Pushing and pulling have both been identified as risk factors in lower back pain (Chaffin *et al.* 1984) and shoulder injury (Hoozemans *et al.* 2004). Pope found 20% of lower back pain problems studied (in the U.S.) is related to pushing/pulling tasks (Pope 1989). Regarding nursing activities carrying and pushing were the only significant tasks associated with back pain (Harber, 1987). The fact that pushing and pulling can lead to injuries in the workplace has led to the development of manual handling guidelines, which will now be discussed.

2.5.3 Manual Handling Guidelines

Snook and Ciriello's (1997) investigation into maximum push and pull forces has been used as the basis for the European Community Directive 90/269, which recommends maximum push and pull force guidelines to prevent injury (Chaffin *et al.* 1984). Their recommended limits are used here to evaluate the forces found in the current study for attendant propulsion of wheelchairs. The guidelines state that when pushing over a distance of 45 m once every 1 minute[10], the initial peak force should not exceed 140 N for males and 120 N for females, the average sustained force should not exceed 70 N for males and 50 N for females.

[10] This is equivalent to 0.75 m/s

	Force [N]		
	Starting	going	stopping
Male	140	70	-140
Female	120	50	-130

Table 2-3: Manual handling guidelines taken from Snook and Ciriello (1997)

2.5.4 Age and push strength

Recent research (Voorbij & Steenbekkers, 2002) has shown it is possible to predict a generalised strength score based on the age of people. The graphs produced by (Voorbij & Steenbekkers 2002) were based on participants completing 4 different tasks namely, pushing, pulling, twisting and gripping. The results of the strength score with respect to age are shown in Table 2-4. The maximum strength for pushing for males was found to be approximately 520 N and for females it was found to be approximately 330N. Interestingly the maximum pull force for males was only in the region of 350N, while females managed 250N. It is not clear what caused the larger drop for males and may be due to differences in stance or technique used.

Age Group	Men (%)	Women (%)
20-30	100	100
50-54	92	91
55-59	94	87
60-64	83	86
65-69	80	77
70-74	72	72
75-79	69	68
80+	62	57

Table 2-4: Percentage decrease in strength with age based on the standardised strength score, taken from Vooribj & Steenbij (2001)

The results of Voorbij (2002) were “remarkably” similar to those of Metter et al (1997) who had carried out a similar smaller study. These results are in contrast to the conclusion of an earlier review of the literature by Damms (1994). The large (n=750) stuffy (Voorbij, 2002) also calculated correlation coefficients between the four tasks the participants had been asked to do. High partial correlation coefficients were found between push strength and pull strength (0.82) and also between push strength and grip strength (0.72); indicating the possibility of using one type of strength test to predict another.

The study by was part of a larger body of work that stemmed from a lack of knowledge for product designers into a variety of human characteristics from the strength score to anthropometric measurements and hand-eye co-ordination. Thus, a standard strength test could be developed to be used during wheelchair prescription for attendants, which could then be used to predict the capacity of the attendant.

The push force capability (provided capability) of people aged between 65 and 69 has been measured by (Steenbekkers & Van Beijsterveldt 1998). This push force test formed a small part of a very large study to measure design relevant characteristics (such as grip strength, push strength, arm reach etc) for aging users of everyday products. The study recommends designing products for the 'weakest individual (or even less)' when designing products that will require push strength from people. They found people aged 65-69 had a mean push force capability of 329 N (n=101), compared with 411 N for those aged 20-30 (n=122). The 5th percentile values for push strength for 65-69 year olds was 157 N, and 128 N for pulling strength (n=128). Further work by this research group has found that the trend of decreasing push strength with age can be predicted using for men (Equation 8) and for women (Equation 9).

$$\mathrm{F} = 116.49 + \frac{395.66}{1 + \left(\frac{\mathrm{A}}{80.78}\right)^{5.71}}$$

Equation 8: Equation for the decrease in push force capability with age for men. F is the push force in Newtons and A is the age in years.

$$\mathrm{F} = -1636.92 + \frac{1978.28}{1 + \left(\frac{\mathrm{A}}{157.14}\right)^{3.75}}$$

Equation 9: Equation for the decrease in push force capability with age for women. F is the push force in Newtons and A is the age in years.

Using Equation 8 and Equation 9, the average capability of a 65 years old male would be 423 N and 65 years old female would be 272 N.

2.5.5 Attendant Propulsion

The only study which focuses to measure the forces involved in everyday attendant wheelchair propulsion was done by van der Woude *et al.* (van der Woude *et al.* 1995). This study investigated the effect of handle height on net joint forces and moments in the upper limbs during attendant wheelchair propulsion whilst carrying out 3 standardised tests; flat pushing, slope pushing and lifting. Each task was completed with the handle height set to 61 %, 69.5 %, 78 % 86.5 % and 95 % of the participant's shoulder height. The horizontal push force was found to range (depending on the handle height) from 94.5 N- 114.1 N when pushing on the flat, which is well below the maximum pushing values found by Steenbekkers & Van Beijsterveldt (see Section 2.5.4), but higher than the current manual handling guidelines (see Section 2.5.3).

In general, higher push handle heights were preferred and a recommendation is made to make handle heights 86.5% of shoulder height. This height showed reduced net moments around the shoulders during flat and inclined pushing; reduced external vertical forces on the hands during flat pushing; reduced net moments around L5-S1 joint; and it allowed attendants to push a wheelchair onto a curb without having to lift it initially, thus reducing net moments on all joints. Incidentally, this is in stark contrast to the findings of Frank and Abel (Abel & Frank 1991) who concluded the magnitude of moments around the shoulder was independent of handle height. However, the methodology used to measure and analyse these moments is unclear.

A carry on study to the above van der Woude study looked at the effect of varying floor surface on rolling resistance of a wheelchair, which found that increasing the coefficient of friction of the floor covering, the rolling resistance increased (van der Woude *et al.* 1995). However, both this study and the original (van der Woude *et al.* 1995) used T-bar shaped handlebars, which is significantly different to the 90 degree handles in a standard 9L wheelchair. Also in both studies, participants were asked to move in the sagittal plane only as a 2D Linked Segment Model was used to model the net forces and moments on the upper body.

Recently ((Jason Tully 2007) addressed the issue of the handle bar design, by carrying out a similar study to that of van der Woude and colleagues (van der Woude *et al.* 1995). Tully instrumented a standard issue 9L wheelchair with uni-dimensional force transducer placed in line with the left push handle and a pair of uni-axial strain gauges positioned on the curve of the left handle. This study built on the van de Woude *et al.*'s methodology by investigating the ground reaction forces (GRF's) of the feet of the attendant as well as the handle forces whilst and attendant pushed a wheelchair with two different occupant weights (75 kg & 95 kg) over two different surfaces (Linoleum and Astroturf) and up a step . He hypothesised that higher ground reaction and handle forces would be found when the occupant weight was increased and also when the frictional properties of the rolling surface were increased. All conditions were compared to the attendants normal gait cycle i.e. walking without pushing a wheelchair.

It was found that the GRF's actually decreased when the rolling resistance of the wheelchair was increased (this was for both an increase in occupant weight and an increase in floor friction (Tully 2007, p.39). One reason given for this was that a proportion of the attendant's weight was being transferred through the wheelchair when pushing it (Tully 2007, p.40). However, this reduction could also have been caused, in part, by a reduced walking speed as the attendants were allowed to self-select their walking speed for all trial conditions. This was investigated by Holloway *et al.* (Holloway *et al.* 2008), who also found the degree of trunk flexion[11] was significantly affected by the occupant weight and the surface.

2.6 Provided Capability: Isometric Push Force

Measuring the maximum isometric push a user is capable of producing has been shown to be correlated to their capability to overcome small gaps (Hashizume *et al.* 2008). They asked users to push as hard as they could on the handrim of a wheelchair, which was attached to a 1-dimensional force transducer, which was fixed to measure forces in the direction of forward movement. This was then normalised for total weight of the user and wheelchair, the result was termed the Normalised Driving Force (NDF) (Equation 10). This measure was found to be correlated to people's ability to overcome small gaps (0mm to 100mm) at varying heights (-20mm to 60mm). In the study it was found that wheelchair users who were able to 'do a wheelie' (be able to lift the casters off of the ground in a

[11] defined as the vector angle between the vertical axis (of the lab frame) and the line connecting the C7 vertebrae and the hip joint centre of rotation, which was estimated as occurring at the head of the greater trochanter

controlled manner) had an NDF score of 0.54 (± 0.08), while those who were unable to complete a wheelie had an average NDF of 0.27 (± 0.12) (Hashizume *et al.* 2008).

$$\mathrm{NDF} = \frac{\text{Maximum Wheelchair Driving Force}}{\text{User'sBody Weight} + \text{Wheelchair Weight}}$$

Equation 10: Equation to calculate the Normalised Driving force developed by Hashizume *et al.* 2008

Elsewhere, the maximum isometric force has been used as an input to the 'Performance:Capacity Ratio' (Equation 11) (Nicholson 2006). This measure is used to see how much of the user's available strength (Capacity) they are using when carrying out a particular task (Performance). It allows for easy comparison between different wheelchair set-ups and also different rolling surfaces and has been used by Hills to quantify the effect of different functional tasks under two different wheelchair set-ups (Hills 2010). A 'functional task' is a phrase commonly used by rehabilitation therapists to describe a task which a person needs to be able to do in order to achieve another objective (function). It usually refers to tasks most adults, without injury would think of as normal e.g. brushing one's teeth or being able to get into and out of bed. In the study by Hills the functional tasks are defined as being able to propel over flat Lino flooring, flat Astroturf, up a 1:12 slope and up a 3 inch kerb. It was found in this study that people often exceeded their Capacity measure when they went up the slope or kerb, and one person exceeded their capacity on the Astroturf.

$$\text{P: C Ratio} = \frac{\text{Peak Mz duing test condition}}{\text{Peak Mz during maximum isometric push}} \text{x } 100$$

Equation 11: Equation to calculate the Performance: Capacity Ratio developed by (Nicholson 2006).

The fact that people exceeded their capacity shows the ratio does not have a defined maximum. The concept of measuring the Capacity someone has to produce the force necessary to accomplish a task is valid, as has been shown by Hashizume *et al.* However, what is also clear is that what both Hills and Hashizume *et al.* are measuring is a voluntary maximum. If a person is scared of tipping backwards, which can be the case when climbing a kerb or a slope, they may well produce more force than they would like to produce. The P:C Ratio was developed with therapists in mind, as a means to make quick and easy comparisons of the effects of changes in wheelchair technique or set-up (Nicholson 2006) and from this point of view the P:C ratio is a valid tool.

The P:C Ratio is similar to Mechanical Use (MU), which uses the resultant force rather than the propulsive moment to define the percentage of a user's force capacity they are using in any given task (Desroches *et al.* 2008; Aissaoui *et al.* 2002). Mechanical Use measures the horizontal force when the wheelchair is restrained (termed the Maximum Voluntary Force, MVF), using a 1-dimensional force transducer attached to the right wheel, it then divides the resultant force calculated from a SmartWheel and multiplies by 100 to get the MU (see Equation 12) (Desroches *et al.* 2008; Aissaoui *et al.* 2002). The measure is averaged over the duration of the push phase of the stroke cycle, which was defined as a 5% increase in Mz from baseline. Deroches *et al.* claim the MU 'is normalised between 0 and 100 '(Desroches *et al.* 2008 pg 1157). However, both studies, using the MU measure were conducted on a treadmill and while theoretically the MU should not exceed 100% it is likely that the metric may suffer from the same problem as the P:C Ratio if tested on more real life conditions such as slopes or kerbs.

$$\mathrm{MU} = \frac{\mathrm{F_{res}}}{\mathrm{MVF}}\mathrm{x}100$$

Equation 11: Equation to calculate Mechanical Use (MU), using the resultant force (F_{res}) during a push compared to the Maximum Voluntary Force (MVF) when the wheelchair is restrained

When pushing on level terrain it can be seen that many people do not utilise all of their available strength with values of MU averaging 17.8% (±12.5%) when recorded under various seating positions by Desroches *et al.* and ranging between 26.6% - 30.3% when seat angles were investigated by Aissaoui *et al.* (Desroches *et al.* 2008; Aissaoui *et al.* 2002). P:C Ratios for flat, steady state lino conditions were also low when measured by Hills for users with low-level injuries, ranging from 7% - 38%. It could be concluded from these results that people only use up to 40% of their available strength when propelling a wheelchair.

2.7 Guidelines for Footways

The most common topological obstacles wheelchair users encounter are curbs and slopes; be they crossfalls intended for drainage or ramps commonly placed as entry access points for wheelchair users into buildings (Kuijer *et al.* 2003).

As was highlighted by (York *et al.* 2007) there is a lack of evidence for the crossfall gradient and kerb heights recommended in DB32 (Department of The Environment 1992). This however, was replaced by the new Manual for Streets (Department for Transport 2007), which gives a more process driven approach to street design and does not give exact guidelines on the kerb heights or crossfalls. It does however inform designers of their need to adhere to the Disability Act (2000) and to consult the guidelines set out in Inclusive Mobility (Department for Transport 2002). These guidelines regard a maximum of 8% (1:12) as a good guideline for maximum slope for small distances (<1m). However, a lesser gradient of 5% is recommended for longer distances with an ideal of less than 2.5% recommended. It notes that these guidelines are not only to ensure greater numbers will be capable of using the footpath, but that there is a risk of injury to the occupant of the wheelchair from toppling if the slope exceeds the maximum guidelines. With regards to crossfalls 2.5% is recommended as an acceptable maximum. However, this is only a guideline and as such can be exceeded.

2.8 Wheelchair Propulsion and Cross-slopes

Despite of the ubiquity of crossfalls in footway construction, there is a lack of evidence for crossfall gradient guidelines, something that has been acknowledged by the Department of Transport in the UK (Department for Transport 2007). This lack of evidence has also been commented upon by the handful of studies which have attempted to assess the impact of crossfall gradients on accessibility (Kockelman *et al.* 2001; Longmuir *et al.* 2003; Vredenburgh *et al.* 2009; Kockelman *et al.* 2002). Both Kockleman *et al.* and Vredenburgh *et al.* conclude that a crossfall gradient of 2.5% guidelines is more than adequate to allow wheelchair users to access the built environment; in fact they argue the guideline is excessively stringent.

Vredenburgh *et al.* state the results of their research ‘do not support’ the Fair Housing Accessibility Guidelines (1991), which currently recommends a 2% crossfall gradient as a maximum, implying they should be increased to 6% as this gradient over the 20ft (6.096m) distance tested was described as requiring only light or very light effort. In addition two thirds of the 43 people tested said they would feel it would not be problematic to travel along a ramp with a crossfall gradient of 6% when it was 78 feet (23.77m). The results of Kocleman *et al.*’s research concludes the maximum crossfall gradient should be increased to

4% in all cases and a 10% gradient is described as being ‘very reasonable’ when the longitudinal slope is less than 5% and where it is challenging to maintain a 4% crossfall gradient (Kockelman *et al.* 2001).

The resulting conclusions from this research are in stark contrast to those reached by more Richter *et al.* and Brubaker (Richter *et al.* 2007a; Brubaker *et al.* 1986). Both of these studies involved only wheelchair users and took a more clinical approach to the problem by investigating the physiological effect (Brubaker *et al.* 1986) and biomechanical effect (Richter *et al.* 2007a) of crossfalls on the wheelchair user. Brubaker *et al.* found in a single case study that there is a 30% increase in net energy cost when propelling over a 2 degree crossfalls when compared to flat terrain (Brubaker *et al.* 1986). While the study by Richter *et al.* found there was a considerable increase in the amount of power required when a 3 degree (≈5%) crossfall was introduced and again when it was increased to 6 degree (≈10%)(Richter *et al.* 2007a). This resulted in an increase in the number of pushes needed to cover the same distance when on a crossfall (Richter *et al.* 2007a). Interestingly, push angle, cadence and self-selected speeds were unaffected by the degree of crossfall. Thus, it would appear that people simply put more pushes into counter the downward turning moment of the crossfall. However, this study was carried out on a treadmill, which meant the distance the wheelchair could roll down the slope was limited by the width of the treadmill, which may have impacted on the number of pushes made by the user. Also, the surface of a treadmill has different properties than a traditional footway. Richter *et al.* recommend investigation of pushrim biomechanics on the upslope as well as the downslope sides of the wheelchair in future studies. They remark that many people appeared to apply braking forces to the upslope side of the wheelchair, though they were unable to quantify this as they only measured downslope biomechanics (Richter *et al.* 2007a).

As was mentioned in section 2.3.4, work per meter has been used by Chesney & Axelson to measure surface firmness (Chesney & Axelson 1996). They also however tested and number of sloped conditions using a plywood ramp which was tilted to have differing crossfalls and longitudinal slopes. They tested crossfall gradients of between 2% and 20% grades with the longitudinal slope fixed at 2%. They also tested 8% longitudinal slopes with 3%, 5% and 8% crossfalls and 14% longitudinal slopes with 5% and 8% crossfalls. Using linear regression they found that the average work per meter on a crossfall could be found with the following equation:

$$y = 2.351x + 26.489$$

Equation 12: Equation to calculate the average work per meter (y) with a crossfall gradient of x, taken from Chesney & Axelson 1996). This equation was found using linear regression with R2=.996

This equation was arrived at using a single person travelling over the 8 test conditions, which were each 2m in length. A selection of the results found by Chesney and Axelson are given in Table 2-5. This study suffers from the fact crossfall was not tested independently of longitudinal slope. However, it has shown that using work to classify surfaces has the advantage of being sensitive to grade changes of a small as 1% (Chesney & Axelson 1996). Although the authors had set out to find an objective measure of surface hardness, it can be argued that they have actually found a way of measuring accessibility as they have proven it is possible to use work to compare both surface hardness and different grades. However, they only used a single person and it is possible that intra-person variability may affect the robustness of this measure.

Table 2-5: Selection of results from Chesney and Axelson's study to measure the work per meter of various surfaces as an objective measure of firmness

Surface Type	Longitudinal slope [%]/Crossfall [%]	Average work per meter [Nm]
Plywood ramp	2/2	31.54 ± 0.48
	2/5	37.91 ± 0.50
	8/5	82.13± 0.87
Accessible carpet	2/0	51.79± 1.19
Hard trail	2/0	32.62± 0.72

2.9 Conclusions

It has been shown that there are a number of interactions which occur between a user, the environment and the wheelchair when pushing along a footway. All of these interactions can be modelled in the Capabilities Model. It can also be concluded that, a wheelchair user, be they the attendant or the occupant, will have to overcome the rolling resistance and the force caused by the downward turning moment of the crossfall in order to push along a standard UK footway. This footway should have a crossfall gradient of no more than 2.5%, however, as it is only a guideline it may be exceeded on occasion.

It would appear that there is a conflict in the literature as to how difficult crossfalls are for wheelchair users and there has been no study conducted on a standard footway surface with crossfalls to assess the forces needed for wheelchair propulsion in a straight line. Furthermore the difficulties in measuring surfaces such as crossfalls, which may require negative forces to be applied to the wheelchair, have been highlighted.

It has been shown that the risk of injuries to wheelchair users is high, in particular when self-propelling and these injuries have been linked to higher push forces. It is possible that a wheelchair user, in countering the effect of the crossfall, may put themselves at further risk of injury. Also, the ability to apply this force may decrease with age, something which is particularly significant for attendant-propulsion. Furthermore, the amount a person allows a wheelchair to run down slope when traversing a crossfall has not been investigated, which would directly affect the necessary width of footways.

3 Self-Propulsion Methods

In Chapter 2 it was shown that the impact of crossfall gradient on the accessibility of footways can be explored using the Capabilities Model. However, to do so it would be necessary to have a more informed picture of the provided capabilities of the Self-Propulsion Wheelchair System (SPWS). This chapter explains the methods used to measure these provided capabilities.

The first section describes the development of the parameters which were measured to depict the provided capabilities. This is done by explaining how a ‘run’ was divided into 3 phases: Starting, going and stopping (section 3.1). Then the provided capabilities measured in each phase are described in 3.2. The types of contacts applied to the wheelchair during the experiments are detailed in 3.3 along with a description of how they tie-in with the idea of a ‘Coping Strategy’. A number of terms needed for ease of defining the subsequent hypotheses are given in Appendix 2:, and then the hypotheses for the provided capabilities are stated in 1.4.

The details of the experiments to assess the hypotheses are then explained. These begin with the ethics approval and inclusion/exclusion criteria for participants in the experiment (section 3.5), followed by details of the equipment used (section 3.6) and a description of the facility where the experiments took place in section 3.7. The protocols for the experiments are then described (section 3.8). Finally the data analysis methods (section 3.9) are detailed along with the statistical tests (section 3.9.4) used for analysis.

3.1 Defining ‘starting’, ‘going’ and ‘stopping’

As discussed 2.4.2.1, analysis of wheelchair propulsion is often divided into the forces required to start a wheelchair and those needed to keep it moving. Normally the starting (or start-up) phase is taken as comprising of the first 3 pushes, and subsequent pushes make what is sometimes called a ‘steady-state’ phase, where the contacts with the handrim appear near uniform (see 2.4.2.1).

The current experiments never reached what could be considered a ‘steady-state’ phase due to the limited length of each run. This was particularly the case on the 2.5% and 4% crossfalls. These conditions presented an added difficulty as on many occasions contacts with the handrim were negative. This meant there would be little sense in taking an average of ‘steady-state’ values. It was also difficult to choose a particular push to represent the ‘steady-state' phase of each crossfall for example the fourth push[12] as this could on occasion be negative and sometimes positive, which would introduce a wide array of variability into the results.

It was therefore decided to define the starting phase as the first push cycle: beginning when the occupant initially made contact with the handrim and ending when they again made contact with the handrim for the second contact. The going phase was defined as starting at the beginning of the second contact and ending at the start of the last positive contact. The stopping phase consisted of the last negative contact made with the handrim. It was defined as starting from the last positive contact (see Figure 3-1). This was done in order to prevent the Matlab script producing an error on occasions when there was no final negative contact. A schematic representation of the tangential push force plotted against time showing the starting, going and stopping phases is given in Figure 3-1.

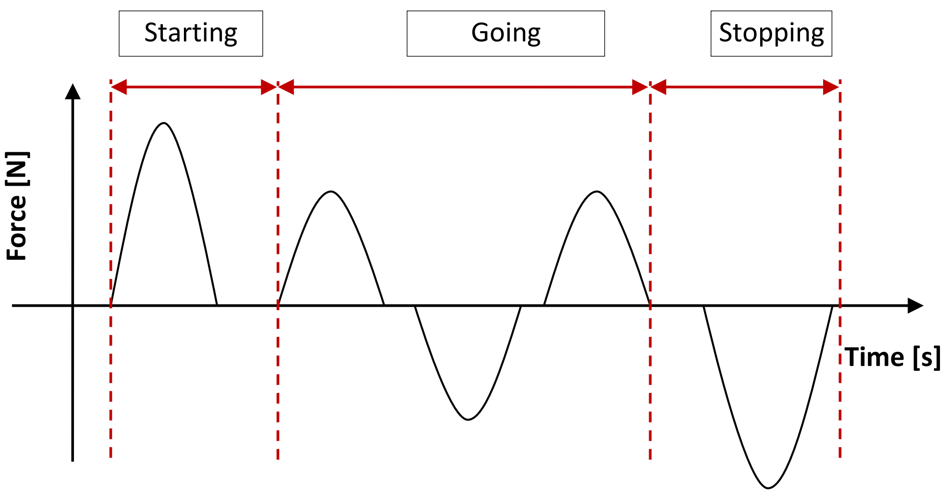


Figure 3-1: Schematic representation of the tangential force applied to a handrim in a typical run, showing the definitions of the Starting, Going and Stopping phases.

3.2 Provided Capabilities on Crossfalls

As was discussed in section 2.2.1 when self-propelling a wheelchair along a footway, the interaction between the Self-Propulsion Wheelchair System (SPWS) and the Environment produce a number of required capabilities which must be overcome in order to be able to traverse the footway in a straight line. These essentially comprise of the rolling resistance

[12] The forth push, being after the initial three which comprise the start-up phase of propulsion is often used as a representative push to describe steady-state conditions e.g. (Cowan et al. 2008)

which exists between the wheelchair and the footway surface and the turning moment caused by the crossfall.

The required capabilities to overcome the combinations of these two forces during the starting, going and stopping phase were measured before the experiments took place. In order to do this an electric scooter pulled the self-propelled wheelchair along footways of 0 %, 2.5 % and 4 % crossfall gradient[13]. The casters of the wheelchair were locked, to prevent the wheelchair from turning downslope. However, it proved impossible to prevent the wheelchair travelling downslope on the 2.5% and 4% crossfalls, due to slipping occurring between the wheelchair castors and wheels and the footway surface. This produced some irregular values in the measured required capabilities. However, it was possible to plot an overall trend for each, the details of these experiments are given in Appendix 3 and a summary of the results for the amount of required going work are given in Table 3-1.

Table 3-1: Mean values of required work for the going phase for each target velocity.

Target Velocity [m/s]	0% Wk_{going} [Nm]	2.5% Wk_{going} [Nm]	4% Wk_{going} [Nm]
0.70	181.58	199.02	229.64
0.81	206.85	226.06	235.27
1.00	207.04	248.28	250.60
1.20	207.24	274.03	282.27
1.45	249.19	237.57	263.68

The focus of this thesis is to investigate the provided capabilities of the Users when faced with a crossfall. Occupants could chose a variety of methods for combating the crossfall, which are discussed in section 1.3, the results of these different methods would need to achieve the following three things in order for them to overcome the required capabilities:

1. They would need to produce the peak force necessary when starting the wheelchair.

[13] In fact the same facility layout and the same wheelchair which are described later in this chapter were used for this test

2. They would need to produce the work necessary to move the wheelchair the required distance during the going phase.
3. They would need to produce the peak force necessary when stopping the wheelchair.

In the case of the SPWS the force which is responsible for the work being done is applied by the occupant to each side of the wheelchair, rather than being pulled by one central point which was the case when measuring the required capabilities (see Appendix 3:). Furthermore the casters will not be fixed in the experiments to measure the provided capabilities, so the wheelchair will be free to turn and travel downslope. The combination of these factors means that the increase in force and work seen in Table 3-1 when the crossfall gradient increases from 0%, will necessitate a difference in the force applied to the upslope and downslope sides of the wheelchair. Therefore, it was thought that when traversing a crossfall there would need to be a difference in force applied to the wheel which was 'upslope' compared to the one which was 'downslope' (these terms will be defined in detail in section 3.7.1).

As the occupant's mass is part of the SPWS which needs to be moved in order to complete the task, it was thought the total force needed to move the wheelchair along with the difference of force would depend on the occupant's mass.

This leads to a number of hypotheses, which are detailed in section 3.4.

3.3 Coping Strategy & Types of Contacts

It was felt important not to ignore the fact people had used 'controlling' contacts to traverse the footway as well as traditional pushes. Therefore, a method needed to be developed which did not dilute the information contained in the irregularity of contacts, but was able to quantify it in some meaningful way.

As the Capabilities Model has the person at its centre, it allows for the different ways in which people decide to tackle a barrier in the environment to be explored. The barrier in this instance is the rolling resistance and the effect of gravity due to the slope and the only method available to the occupant to overcome these forces is to apply force to the handrim. Therefore, the way in which people adapt to the changing crossfall, or put another way their 'coping strategy' must come from the way in which they impart force to the handrim. For this reason within each going phase the contacts were divided into pushes, brakes and impacts. The latter two types are both seen as ‘controlling’ contacts, but are distinct from

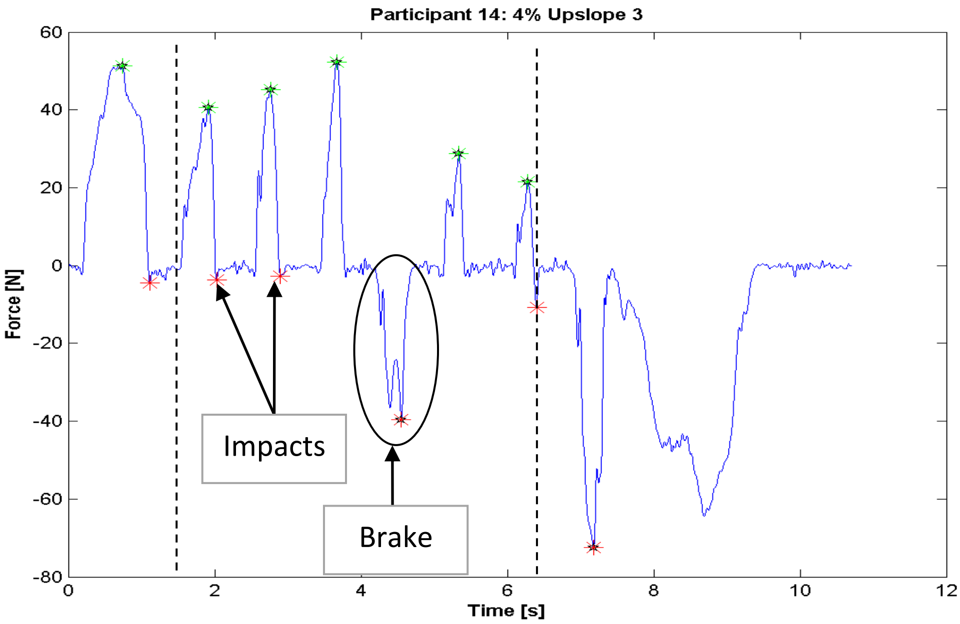


Figure 3-2: Sample tangential force plot against time with ‘Impacts’ and ‘Brakes’ which occur in the Going phase highlighted. Green stars represent the peak push forces, red stars the impact peaks and the red-black stars the brakes. The vertical dashed lines represent the divisions between the starting, going and stopping phases.

each other in that the brakes appear in lieu of pushes whilst the impacts occur when the hand initially makes contact with the handrim. These Impacts are akin to Kwarciak *et al.*’s (2009) definition of ‘initial contact’ when they occur immediately before the push and ‘release’ when they occur immediately after a push. Figure 3-2 highlights the difference between the Impacts and Brakes.

Each contact needed to be described. To do this the key contact parameters of Peak Tangential Force (F_{tPK}), Contact Time ($T_{contact}$) and contact frequency were calculated. These parameters were chosen as Ft_{PK} and $T_{contact}$ were thought to be the primary descriptors of a contact, as with these two parameters a crude plot of the contact force can be made either with a triangle or a more complete description made using a sine function (see Figure 3-3). Therefore with the frequency of each contact type a complete description of the coping strategy can be described. How each contact was found and their precise definitions are given in Section 3.3 .

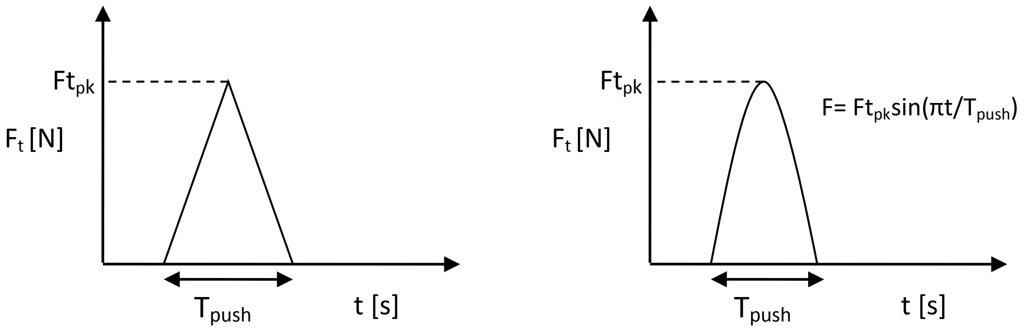


Figure 3-3: Representative curves of the tangential force (Ft) against time (t) constructed using Peak tangential force and push time (Tpush). Left shows a simple isosceles triangle function. Right shows a sine function.

3.4 Objectives and Hypotheses

The aim of this thesis I to measure the effect of crossfalls on the provided capabilities needed to push a wheelchair along a footway. At the top level if a person is unable to start the wheelchair, keep the wheelchairs moving in a straight line or stop the wheelchair then the required capabilities will have exceed the provided capabilities. When there is no crossfall present this gives rise to 3 provided capabilities: the capability to apply sufficient force to overcome the static friction of the wheelchair system, the capability to produce sufficient work to keep the wheelchair system moving in a straight line and the capability to stop the wheelchair system (see Figure 3-4). When there is a positive crossfall present there will need to be a difference of force applied to the upslope and downslope sides of the wheelchair when starting, going and stopping. Therefore, it is hypothesised that an additional number of provided capabilities are needed when pushing on a positive crossfall. To test this hypothesis key parameters have been chosen for each section of a run (see Figure 3-4).

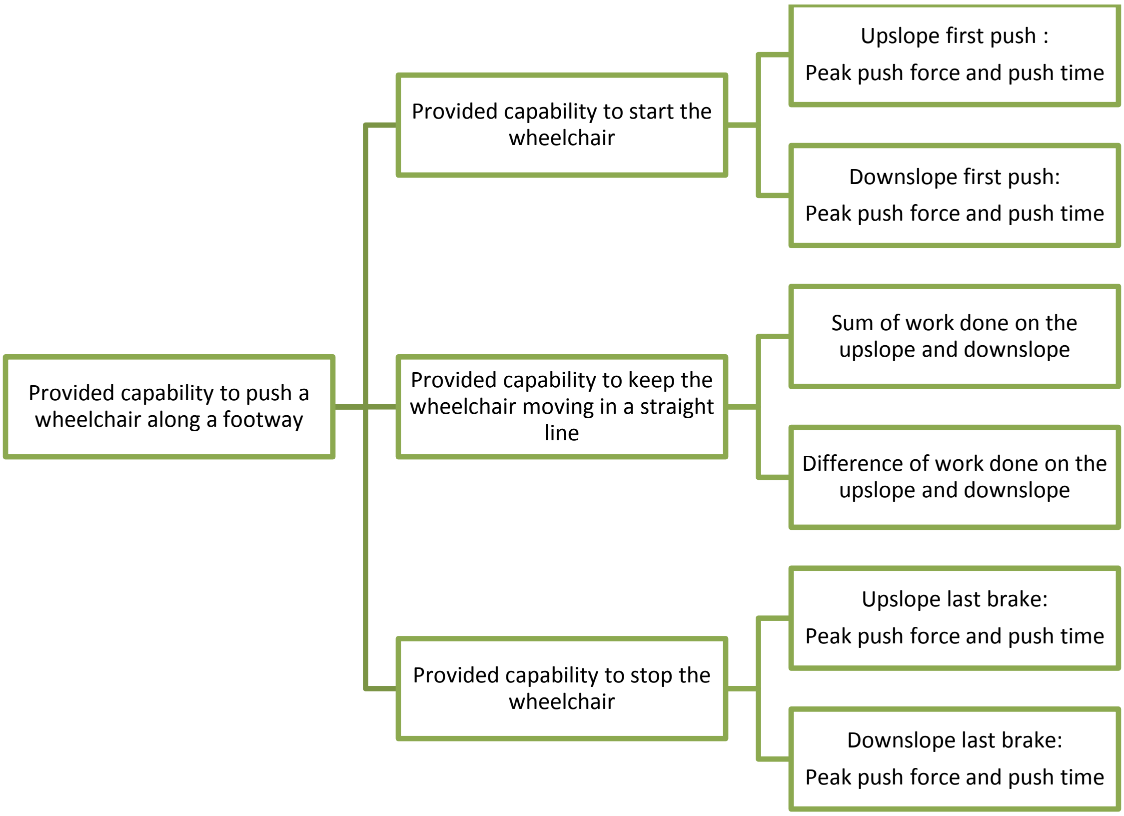


Figure 3-4: Provided capabilities at each phase of a run along with key variables

When starting and stopping the peak tangential force and time of contact for the first push are taken as the key variables describing the provided capability of the user. These variables give rise to the specific hypotheses given in 3.4.1, which test for the effect of crossfall gradient and occupant weight on both the upslope and downslope sides of the wheelchair.

During the going phase difference of work and sum of work are chosen as the key variables. These give rise to the specific hypotheses described in 3.4.2. These again test for the effect of crossfall gradient and occupant weight on the key parameters.

In order to concisely specify the hypotheses a number of terms have been defined, which can be found in Appendix 2:. These terms have their first letters capitalised in order to aid the reader.

3.4.1 Starting & stopping Phases

3.4.1.1 Downslope

Hypothesis 1 and Hypothesis 2 test the effect of crossfall gradient on the provided capabilities of the SPWS for the downslope side of the wheelchair during the starting phase.

Hypothesis 1.

H0: There will be no change in the Provided Starting Force regardless of crossfall gradient on the downslope side.

H1: There will be a significant increase in the Provided Starting Force as crossfall gradient increases on the downslope side.

Hypothesis 2.

H0: There will be no change in the Start Push Time regardless of crossfall gradient on the downslope side.

H1: There will be a significant increase in the Start Push Time as crossfall gradient increases on the downslope side.

Hypothesis 3 and Hypothesis 4 test the effect of crossfall gradient on the provided capabilities of the SPWS for the downslope side of the wheelchair during the stopping phase.

Hypothesis 3.

H0: There will be no change in the Provided Stopping Force regardless of crossfall gradient on the downslope side.

H1: There will be a significant decrease in the Provided Stopping Force as crossfall gradient increases on the downslope side

Hypothesis 4.

H0: There will be no change in the Stop Push Time regardless of crossfall gradient on the downslope side.

H1: There will be a significant decrease in the Stop Push Time as crossfall gradient increases on the downslope side.

3.4.1.2 Upslope

Hypothesis 5 and Hypothesis 6 test the effect of crossfall gradient on the provided capabilities of the SPWS for the upslope side of the wheelchair during the starting phase.

Hypothesis 5.

H0: There will be no change in the provided starting Force regardless of crossfall gradient on the upslope side.

H1: There will be a significant increase in the provided Starting Force as crossfall gradient increases on the upslope side.

Hypothesis 6.

H0: There will be no change in the Start Push Time regardless of crossfall gradient on the upslope side.

H1: There will be a significant increase in the Start Push Time as crossfall gradient increases on the upslope side.

Hypothesis 7 and Hypothesis 8 test the effect of crossfall gradient on the provided capabilities of the SPWS for the upslope side of the wheelchair during the stopping phase.

Hypothesis 7.

H0: There will be no change in the Provided Stopping Force regardless of crossfall gradient on the upslope side.

H1: There will be a significant decrease in the Provided Stopping Force as crossfall gradient increases on the upslope side

Hypothesis 8.

H0: There will be no change in the Stop Push Time regardless of crossfall gradient on the upslope side.

H1: There will be a significant decrease in the Stop Push Time as crossfall gradient increases on the upslope side.

3.4.2 Going Work and Going Work Difference

Hypothesis 9 and Hypothesis 10 test the effect of crossfall gradient on the provided capabilities of the SPWS for the going phase.

Hypothesis 9.

H0: There will be no change in the Provided Going Work regardless of crossfall gradient.

H1: There will be a significant increase in the Provided Going Work as crossfall gradient increases.

Hypothesis 10.

H0: There will be no change in the Provided Going Work Difference regardless of crossfall gradient.

H1: There will be a significant increase in the Provided Going Work Difference as crossfall gradient increases.

Hypothesis 11 and Hypothesis 12 test the effect of occupant mass on the provided capabilities of the SPWS for the going phase.

Hypothesis 11.

H0: There will be no change in the Provided Going Work regardless of occupant mass.

H1: There will be a significant increase in the Provided Going Work as occupant mass increases.

Hypothesis 12.

H0: There will be no change in the Provided Going Work Difference regardless of occupant mass.

H1: There will be a significant increase in the Provided Going Work Difference as occupant mass increases.

If a significant difference is found for any of the hypotheses related to provided going work difference, it is possible that the difference was the result of a change to the forces, and in turn the work, done on the upslope and/or the downslope side of the wheelchair. For this reason, when significant results occur, they will be followed by a formal test on the upslope and downslope parameter. These hypotheses are not stated here, but will be stated as and when they are tested in section 5.5.2.

3.4.3 Coping Strategy

In this thesis three types of contacts have been identified in the going phase, these are: Pushes, Brakes and Impacts. In order to cope with the change in crossfall the occupant can chose to change the quantity of each type of contact. They can also choose to change the magnitude or duration of each contact type. The background to these parameters is given in section 3.3. These strategies were thought to change differently depending on the side of the wheelchair being investigated.

These possible strategies give rise to a set of hypotheses for the downslope side and a separate set for the upslope side.

3.4.3.1 Downslope

Hypothesis 13, Hypothesis 14 and Hypothesis 15 test the number, peak magnitude and duration of Pushes on the Downslope side of the wheelchair.

Hypothesis 13.

H0: There will be no change in the number of pushes in the going phase regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant increase in the number of pushes in the going phase as crossfall gradient increases on the downslope side of the wheelchair.

Hypothesis 14.

H0: There will be no change in the magnitude of the average push F_{tPK} regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant increase in the magnitude of the average push F_{tPK} as crossfall gradient increases on the downslope side of the wheelchair.

Hypothesis 15.

H0: There will be no change in the average duration of push regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant increase in the average duration of push as crossfall gradient increases on the downslope side of the wheelchair.

Hypothesis 16, Hypothesis 17 and Hypothesis 18 test the number, peak magnitude and duration of Brakes on the Downslope side of the wheelchair.

<u>Hypothesis 16.</u>

H0: There will be no change in the number of brakes in the going phase regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant decrease in the number of brakes in the going phase as crossfall gradient increases on the downslope side of the wheelchair.

<u>Hypothesis 17.</u>

H0: There will be no change in the magnitude of the average brake F_{tPK} regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant decrease in the magnitude (absolute value) of the average brake F_{tPK} as crossfall gradient increases on the downslope side of the wheelchair.

<u>Hypothesis 18.</u>

H0: There will be no change in the average duration of a brake regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant decrease in the average duration of a brake as crossfall gradient increases on the downslope side of the wheelchair.

Hypothesis 19, Hypothesis 20 and Hypothesis 21 test the number, peak magnitude and duration of <u>Impacts</u> on the <u>Downslope</u> side of the wheelchair.

<u>Hypothesis 19.</u>

H0: There will be no change in the number of impacts in the going phase regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant decrease in the number of impacts in the going phase as crossfall gradient increases on the downslope side of the wheelchair.

<u>Hypothesis 20.</u>

H0: There will be no change in the magnitude of the average impact F_{tPK} regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant decrease in the magnitude (absolute value) of the average impact F_{tPK} as crossfall gradient increases on the downslope side of the wheelchair.

Hypothesis 21.

H0: There will be no change in the average duration of an impact regardless of crossfall gradient on the downslope side of the wheelchair.

H1: There will be a significant decrease in the average duration of an impact as crossfall gradient increases on the downslope side of the wheelchair.

3.4.3.2 Upslope

Hypothesis 22,Hypothesis 23 and Hypothesis 24 test the number, peak magnitude and duration of Pushes on the Upslope side of the wheelchair.

Hypothesis 22.

H0: There will be no change in the number of pushes in the going phase regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant decrease in the number of pushes in the going phase as crossfall gradient increases on the upslope side of the wheelchair.

Hypothesis 23.

H0: There will be no change in the magnitude of the average push F_{tPK} regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant decrease in the magnitude of the average push F_{tPK} as crossfall gradient increases on the upslope side of the wheelchair.

Hypothesis 24.

H0: There will be no change in the average duration of push regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant decrease in the average duration of push as crossfall gradient increases on the upslope side of the wheelchair.

Hypothesis 25,Hypothesis 26 and Hypothesis 27 test the number, peak magnitude and duration of Brakes on the Upslope side of the wheelchair.

Hypothesis 25.

H0: There will be no change in the number of brakes in the going phase regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant increase in the number of brakes in the going phase as crossfall gradient increases on the upslope side of the wheelchair.

Hypothesis 26.

H0: There will be no change in the magnitude of the average brake F_{tPK} regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant increase in the magnitude of the average brake F_{tPK} as crossfall gradient increases on the upslope side of the wheelchair.

Hypothesis 27.

H0: There will be no change in the average duration of a brake regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant decrease in the average duration of a brake as crossfall gradient increases on the upslope side of the wheelchair.

Hypothesis 28,Hypothesis 29 and Hypothesis 30 test the number, peak magnitude and duration of Impacts on the Upslope side of the wheelchair.

Hypothesis 28.

H0: There will be no change in the number of impacts in the going phase regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant increase in the number of impacts in the going phase as crossfall gradient increases on the upslope side of the wheelchair.

Hypothesis 29.

H0: There will be no change in the magnitude of the average impact F_{tPK} regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant increase in the magnitude of the average impact F_{tPK} as crossfall gradient increases on the upslope side of the wheelchair.

<u>Hypothesis 30.</u>

H0: There will be no change in the average duration of an impact regardless of crossfall gradient on the upslope side of the wheelchair.

H1: There will be a significant decrease in the average duration of an impact as crossfall gradient increases on the upslope side of the wheelchair.

3.5 Ethics

The experiments were approved by the Ethics Committee at University College London. People were eligible for recruitment if they were between the ages of 18-65, had no history of shoulder injury and who felt comfortable propelling along a 10m pavement, with a crossfall gradient, in a light weight wheelchair. Wheelchair users were recruited if they were regular wheelchair users. Non-wheelchair users were recruited if they had no experience of using a wheelchair. Participants were recruited via an email that was sent out to all students in the Civil Environmental and Geomatic Engineering Department. Individual wheelchair users who had previously taken part in experiments for the Accessibility Research Group (ARG), and who had given permission to be contacted by ARG with details of upcoming experiments were also contacted.

An information sheet was provided to all participants, with slightly different wording for wheelchair users than for non-wheelchair users and written consent was gained before the experiments took place.

3.6 Equipment

The equipment used in these experiments will now be described, beginning with the type of video recording equipment used (section 3.6.1). The wheelchair used is then described along with its set-up in section 3.6.2. The SmartWheel is described in section 3.6.3 and the SmartWheel data files which were used in this study are detailed in section 3.6.4.

3.6.1 Video Recording System

The experiments were videoed using 4 different camera angles and recorded using a system called Canopus EMR100 with accompanying Mediacruise software version number 2.205. This system scans in the CCTV footage and records Mpeg files which can contain 1 or more video streams. The parameters used in this study are given in Table 3-2. Three snapshots of the types of video angles are shown in Figure 3-5.

Parameter Name	Parameter Value
Mode	Mpeg1
Resolution	352 x 288
Standard	PAL
Sample Rate	48 Hz

Table 3-2: Video recording parameters

3.6.2 The Wheelchair

The chosen wheelchair was a 17” x 17” (40.64cm x 40.64cm) Quickie GPV, rigid frame lightweight wheelchair with a 3” (7.62cm) 'Optimus cushion (see Figure 3-6). The wheelchair was chosen as it is a common wheelchair prescribed to active users by the NHS. The seat had a 4cm bucket, meaning there was a 4cm height difference between the front of the seat (45cm) and the rear of the seat (41cm). The wheelchair was fitted with 25” (.635 m) solid tyre on one side and the SmartWheel on the opposing side. The wheelbase of the

Figure 3-4: Example camera angles snapshots. Left shows birds-eye view on 0%, centre shows the 'fishbowl' view and the right snapshot shows the elevated overview.

wheelchair was 41 cm and the rear wheels were set-up with a 2 ° camber. The casters of the wheelchair were solid and 5” (.127 m) diameter.

Figure 3-6: Quickie GTX Wheelchair used in this study

3.6.3 The SmartWheel

The SmartWheel (SW)[14] is a commercially available wheel, which attaches to the axle receiver of a wheelchair in place of a standard wheel. It is capable of measuring the three dimensional forces and moments applied to its handrim, as well as the velocity of the wheelchair (see Figure 3-7).

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 *et al.* 1997).

The sign convention was used for these experiments was the same as is defined in the SmartWheel User Manual (Three Rivers 2006) ; F_x is positive when going forward/anterior, F_y positive when perpendicular to the ground/superior and F_z positive when pointing towards the centre of the wheelchair/medial when the wheelchair was on the left side and pointing away from the wheelchair/lateral when on the right side (see Figure 3-7) . The moments (M_x, M_y and M_z) are defined as positive when rotating counter clockwise about the respective axis.

[14] The SmartWheel is developed by Three Rivers Holdings, LLC.

The total force (F_{tot}) is then calculated using $F_{tot} = \sqrt{F_x^2 + F_y^2 + F_z^2}$ and the tangential force (F_t) is calculated using $F_t = \frac{M_z}{r}$, where M_z is the moment about the rear wheel axle (see Figure 3-7) and r is the radius if the rear wheel[15].

When collecting data on the SmartWheel it is necessary to use the SmartWheel software. This can be set to collect data in 'Research Mode' or 'Clinical Mode'. For these experiments the data was collected in Research Mode.

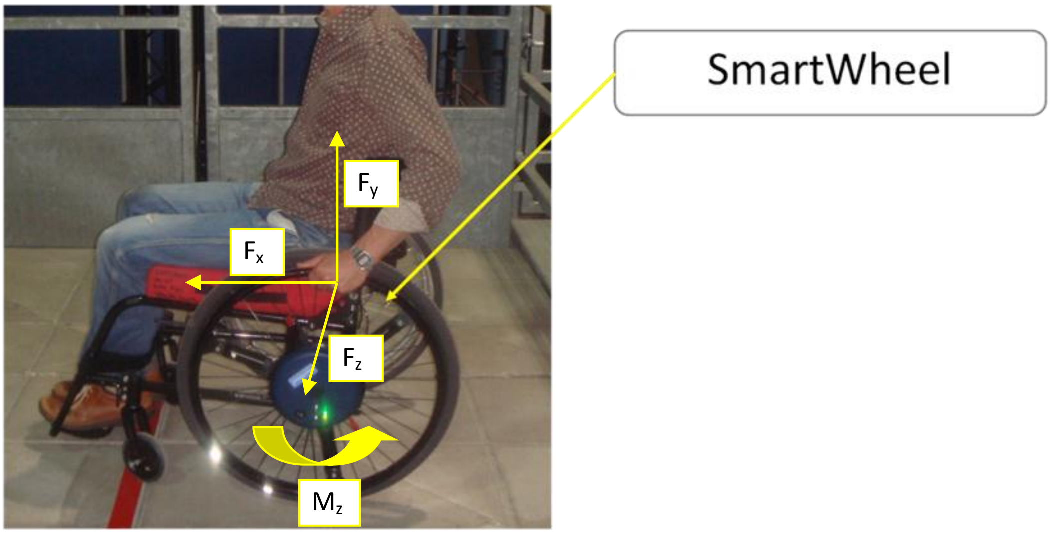


Figure 3-5: Picture of wheelchair and SmartWheel used in the study showing the positive directions of the forces along the three orthogonal axes (Fx, Fy and Fz) along with the moment about the z axis (Mz)

3.6.4 SmartWheel Data Files

As has already been mentioned in section 2.4.1 the SmartWheel is capable of measuring 3D forces and moments as well as velocity. The data is recorded on an SD memory card and is also transmitted wirelessly to a PC. Recorded data files are saved as comma separated files, which can be imported directly into programs such as Microsoft Excel or Matlab.

The SmartWheel also comes with a data analyser package of its own called the Data Analyzer Tool. The Data Analyzer Tool used in this study was the version which is incorporated into the SmartWheel Software, 2006, v.1.6.0. A description of how to use this tool and the files it generates is given in the SmartWheel User's Guide (Three Rivers 2006). The file type used in this study was 'Format 2', which contains the following variables: wheel angle, speed, distance, F_x, F_y, F_z, M_x, M_y, M_z, F_t and F_{tot} (see Figure 3-8).

[15] *This definition of F_t assumes that the moment exerted by the wrist is negligible.*

	A	B	C	D	E	F	G	H	I	J	K
1	Angle [deg]	Speed [m/s]	Distance [m]	Fx [N]	Fy [N]	Fz [N]	Mx [N*m]	My [N*m]	Mz [N*m]	Ft [N]	Ftot [N]
2	358.07	0	0	2.38	-7.93	0.1	0.46	-0.17	1.28	4.99	8.28
3	358.07	0	0	2.28	-7.76	0.41	0.51	-0.2	1.36	5.28	8.1
4	358.07	0	0	2.18	-7.82	0.51	0.61	-0.29	1.46	5.67	8.13

Figure 3-8: Screenshot of 'Format 2' type file produced by the Data Analyzer Tool

To create this type of file the Data Analyzer Tool automatically filters and converts the raw data into force and moment data about the three orthogonal global axes x, y and z. This creates F_x, F_y and F_z for the forces acting along the x-axis, y-axis and z-axis respectively.

All variables are calculated using filtered data. The filter used is a digital Finite impulse Response (FIR) filter with a filter length of 32 (Three Rivers 2006) Pg 2-18). There is a 16 sample delay in all calculated variables except the stroke angle (Three Rivers 2006) Pg 2-18) which has not been accounted for in this file type. However, this was corrected for when the files were imported into Matlab for further analysis, which will be described in 3.9.3.

3.7 Facility & Layout

The PAMELA facility allows the reconstruction of real-life street conditions inside a laboratory environment. It contains a platform consisting of 57 square modules which can be tilted in different orientations to represent slopes and crossfalls. They can also be raised to different heights so that steps and kerbs can be reproduced. A detailed description of the PAMELA facility is given by Childs et al (Childs *et al.* 2007).

Each module is 1.2 metres square and the modules themselves can be moved to make various shaped configurations such as a square, L-shape or U-shape. The surface of the modules can also be changed. As the platform is inside the conditions on the platform are unaffected by weather conditions and the lighting conditions can be controlled to a high degree of accuracy, which makes for a highly repeatable environment for testing street layouts.

For this experiment the PAMELA platform was set up so that there were three lanes with different crossfall gradients; 0%, 2.5% and 4%.The lanes were each 2.4m wide and 10.8 m long (see Figure 3-9).

he surface of the platform consisted of concrete pavers, commonly found on UK footways. The lanes were each marked with 2 red tape lines to indicate the start and finish lines. There was also a red dashed line indicating the straight path the participants should attempt to

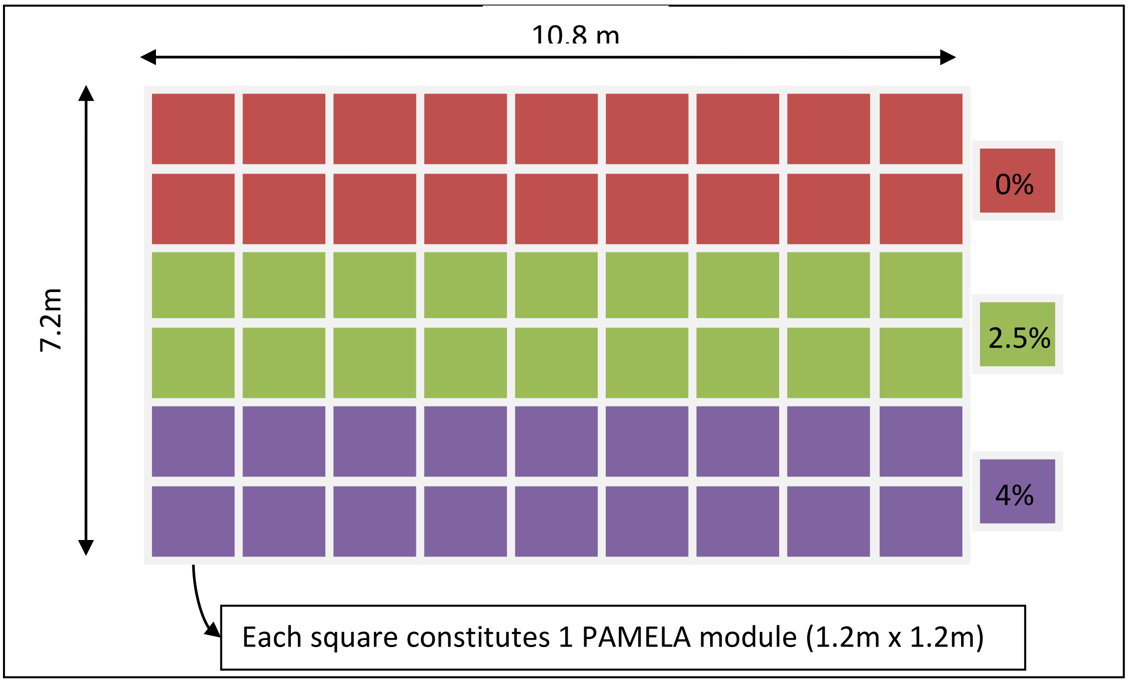


Figure 3-6: Birdseye view schematic of the PAMELA set-up

follow. This was positioned slightly off-centre so that they were nearer the top of the crossfall lane (see Figure 3-10).

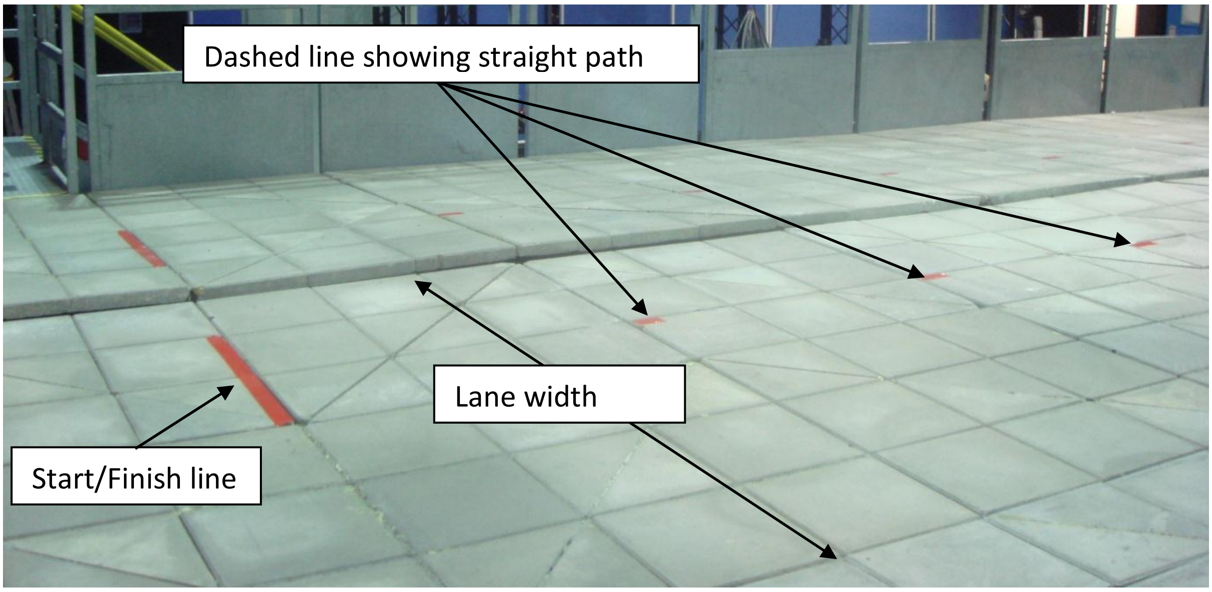


Figure 3-10: Illustrated photo of the PAMELA set-up showing start/finish line and position of dashed line

3.7.1 Upslope and downslope

The study only had one SmartWheel available, therefore when travelling on a sloped surface the SmartWheel was either upslope or downslope (see Appendix 2: for an illustration of these terms). The orientation along with the crossfall direction is shown in Figure 3-11.

For the 0% condition, as there was no difference between North and South runs, the runs

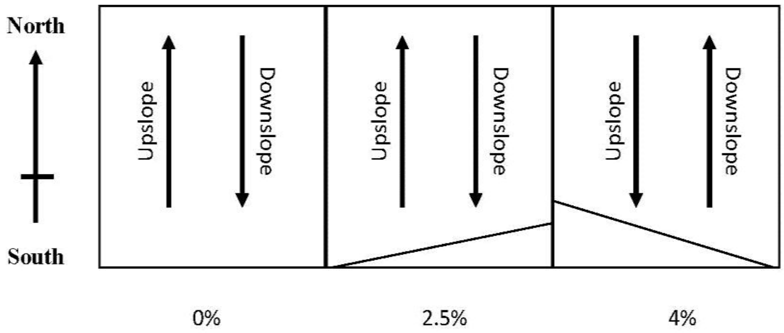


Figure 3-7: Picture describing upslope and downslope lane conditions when the occupant was right handed and so the SmartWheel was on the left hand side of the wheelchair

were arbitrarily assigned to upslope (North) and downslope (South) for the purposes of analysis (see Figure 3-11).

The SmartWheel was placed on the wheelchair user's non-dominant side of the wheelchair i.e. for right handed people the SmartWheel was attached to the left hand side of the wheelchair.

3.8 Protocol

3.8.1 Maximum Voluntary Push Test

A Maximum Voluntary Push Test (MVPF) was completed by each participant to capture the maximum amount of force the participant was capable of applying to the push rim when the wheelchair is restrained from moving.

For the MVPF the wheelchair was placed up against the parapet at the side of the PAMELA platform and the brakes were applied to the wheelchair to prevent it from moving.

It is essential that no force be applied to the handrim when the SmartWheel is initially turned on, and that the SmartWheel not be moving, as a calibration process occurs at this point. For this reason the occupant was asked to place their hands on their lap and sit as still as possible while the SmartWheel was activated.

Once the calibration procedure had been completed the participants were asked to place their hands on the handrim as if they were about to push the wheelchair. They were then given the following verbal instructions:

"When I tell you to 'GO' I want you to push your wheelchair as hard as you can 3 times. The wheelchair should not move. Please push for a count of 3 seconds with a rest of 5 seconds between each push. I will count and time you. When you have finished the 3 second push please remove your hands from the wheelchair. Do you have any questions?" PAUSE "GO."

The SmartWheel recorded the force and moment data from the 3 maximum pushes in a single file. The sampling frequency used was 240 Hz, which is the standard sampling frequency of the SmartWheel.

3.8.2 Crossfall Experiments

The participant was asked to position the wheelchair in the correct starting position. For this to be the case the wheelchair needed to be directly behind the start line, the casters and the wheelchair had to be parallel to the intended direction of travel, the casters must be trailing backwards and the dashed red line had to be mid-distance between the two casters. On occasion help was given to ensure the casters were orientated in the right direction and the wheelchair was in the correct starting position.

The order of conditions was randomised and the run order is given in Table . The randomisation was done by pulling numbered pieces of paper from a container.

Table 3-3: Run order of experiments

Participant Number	Run order		
	1st	2nd	3rd
1	2.5	0	4
2	2.5	4	0
3	2.5	4	0
4	0	2.5	4
5	4	2.5	0
6	2.5	0	4
7	2.5	4	0
8	0	2.5	4
9	4	0	2.5
10	2.5	0	4
11	0	4	2.5
12	2.5	0	4
13	2.5	4	0
14	0	2.5	4

Participants were asked to sit with their hands off the handrim before each trial. This was done to ensure all force data collected related to the run.

Participants were then given the following verbal instructions before completing each for the 3 test conditions:

"When I tell you to 'GO' I want you to push the wheelchair in a straight line by attempting to follow the dashed red line on the floor. Push at a comfortable speed, as if you were pushing on a path. Keep pushing until you pass the stop line. Then stop as quickly as you are able and do not turn the wheelchair. Do you have any questions?" PAUSE " GO."

The SmartWheel software was then set to record and on completion of each trial the SmartWheel files were saved while the participant was stationary. The participant was then asked to position themselves for the next trial, with help given as required.

3.9 Data Analysis Methods

3.9.1 Maximum Voluntary Push Test data reduction

The format 2 files created by the SmartWheel data analyser tool (see section 3.6.4 for more details) for each MVPF were imported into Microsoft Excel where the maximum value of F_t

was found for each set of maximum pushes. The value of F_{tot} at the same time as the maximum F_t was also recorded.

3.9.2 Analysis of deviation from a straight line: Video Analysis

The videos were observed using Windows Media Player V12, and the maximum deviation from a straight line was recorded for each experimental run. Deviation from a straight line was analysed by stepping through the videos frame by frame and measuring the X and y coordinates of the position of the right rear wheel every 0.4m. The analysis was done in 0.4m intervals and the wheel had to completely pass over one paver to be counted as having deviated that 0.4m section. Therefore, deviations less than 400m in 400mm length are not counted.

3.9.3 Data Analysis Methods: provided capabilities

A custom Matlab script was written to analyse the data files produced by the SmartWheel data analyser tool (see section 3.6.4 for more details) for the crossfall data. This script first removed the effect of the 16 sample in all variables except stroke angle. An eighth order low-pass Butterworth filter was then applied with a 20 Hz cut-off frequency, which is the same filter used by Kwarciak et al (Kwarciak *et al.* 2009).

Each contact was then identified using the following criteria. A push was defined as a positive tangential force applied to the handrim for a period of 0.2 seconds or more and a brake was similarly a negative tangential force applied to the handrim for 0.2 second or more. Initially contacts had been defined as occurring for 0.1 seconds, as has been used in the past by Cowan et al . However, this definition led to the inclusion of a number of brakes, which would more aptly be defined as an impact as the user grabs the wheel. Therefore, 0.2 seconds was chosen as a revised threshold.

However, it was clear when looking at the plots of each run that a number of the brakes were in fact part of the subsequent push, and though they add to the work being done on the system, they are different to a brake as they are not performed consciously by the user. To attempt to separate the brakes which could be considered a brake and those which could be considered more of a push 'impulse' the following was done. The contact time of all pushes and brakes were found. These were then separated into downslope brakes, downslope pushes, upslope pushes and upslope brakes. Each group was explored using PASW 18.

The search function Peakdetect [16]was used to apply an inside out search through the F_t data and identify the maximum and minimums along the force trace.

The time and value of the maxima and minima were recorded. The maximum values were then used to identify the start and finish of each contact. This was done by stepping through F_t from the maximum position until the value of F_t fell below the cut-off (set to 0 N). In a similar fashion the minimum values were used to find the points where F_t rose above the cut-off (set to 0 N). Contacts that were shorter than 0.1 seconds were deleted from the list of contacts.

Impulses were found by calculating the difference between the end time and the start time of each consecutive contact. When the start time of the second contact was the same as the end time of the first contact, an Impulse had been detected. The impulse could be due to the 'initial contact' or the release of the handrim. If due to the initial contact then the Impulse would have been the first contact, with the second contact constituting the push. If the impact was due to the release of the handrim then the Impulse would be the second contact and the first the push. Therefore the magnitudes of the two Ft peaks of the compared contacts were examined and whichever was less than zero was determined to be the impact. Using this method allowed for impacts due to 'initial contact' and those due to releasing the handrim to be detected. This method assumes the pushes were the intended form of contact.

Therefore, within this study a 'push' is defined as a positive tangential force applied to the handrim, which occurs for at least 0.1 seconds, which is similar to that used by Cowan et al, with the only difference being that Cowan used the moment around the z-axis rather than the tangential force (Cowan *et al.* 2009). A 'brake' is defined similarly as a negative tangential force applied to the handrim, which occurs for at least 0.1 seconds. Impacts are defined as negative tangential forces which occur immediately after or before a push.

In summary, for each contact the peak F_t, start time, end time, average velocity, peak velocity and work done were calculated. The peak F_t corresponded to the initial maximum

[16] Full details of this function can be found at http://www.billauer.co.il/peakdet.html

or minimum found by the peakdetect function. The start and end times had been found by determining when Ft first dropped below zero on either side of the peak value. The work done was calculated using the trapz function in Matlab, which using the trapezium rule to integrate a series of data. In this case F_t was integrated with respect to distance travelled. Ft, velocity and distance are direct outputs of the SmartWheel.

3.9.4 Statistical Analysis

All statistical analysis was carried out using PASW Statistics 18, Release Version 18.0.0 (SPSS Inc., 2009, Chicago, IL, www.spss.com).

Where data sets were normally distributed multiple linear regression analyses were run. Each parameter (e.g. Provided Start Force) was defined as the independent variable. Crossfall gradient (C) and participant mass (m) were defined as the dependant variables (also termed regressors). Crossfall and weight were coded as continuous variables (i.e. 100Kg would be exactly double 50Kg).

This analysis produces a prediction equation based on the regression coefficients (β) and the independent variables (Tabachnick 2001).

$$P = A + \beta_c C + \beta_w W$$

Equation 13: prediction equation for parameter (P) when dependent variables are crossfall gradient (C), and participant weight (W). A is the constant term.

Before completing the analysis all parameters were checked using a t-test to see if there was a significant effect of trial. With the significance level set to $p=0.05$, there was not. Therefore, it can be concluded there was neither a learning nor a fatigue effect over the course of the experiments and there was no need to add 'Trial' as an independent variable in the regression analyses.

When the data was not normally distributed non-parametric test were carried out. The main alternative test to the multiple regression analysis was the Friedman Test, with post-hoc analysis carried out using Wilcoxon Signed-Rank Test.

A Bonferroni adjustment was applied to the significance level, which resulted in a significance level of $p=.017$. A Bonferroni adjustment is used to correct for the increased chance of accepting a null hypotheses when it should in fact be rejected (a Type 1 error) when making multiple comparisons. To apply the Bonferroni adjustment the original alpha level (in this case p= .05) is divided by the number of planned comparisons. In this case 3 comparisons are being made (0 % against 2.5%, 2.5% against 4% and 0% against d 4%) (Pallant 2005).

4 Attendant-Propulsion Methods

The Methods used to collect the data to measure the provided capabilities of the Attendant-Propulsion Wheelchair System (APWS) will be detailed in this chapter. They are similar to those described in chapter 3 to measure the provided capabilities of the Self-Propelled Wheelchair System. However they differ both in equipment and analysis methods due to the fact it is the Attendant which is providing the force necessary to move the wheelchair, and this is done by applying force to the handles of the wheelchair.

This chapter follows a slightly different structure to chapter 3. The first section describes the development of the parameters which were measured to describe the APSW provided capabilities. Again each run was divided into: starting, going and stopping (Section 4.1). Then the provided capabilities measured in each phase are described in Section 4.2. The hypotheses are then given in Section 4.3.

The experimental methods are then detailed, starting with the ethics approval and inclusion/exclusion criteria for participants in the experiment (Section 4.4), followed by details of the equipment used (Section 4.5). The protocols for the experiments are then described (Section 4.6). Finally the data analysis methods (Section 4.7) are detailed along with the statistical tests (Section 4.7.3) used for analysis.

4.1 Defining ‘starting’, ‘going’ and ‘stopping’

The definitions of starting, going and stopping for the attendant-propelled runs differ somewhat from those defined in section 3.1 for the experiments to find the provided capabilities of the SPWS. This was necessary due to the nature of force application by the attendant fundamentally differing to that of the occupant, something which has been discussed previously.

The starting phase is defined as the time from which force is first applied to one of the handles by the Attendant, to the time of the first local minimum. The going phase begins when the starting phase ends and it finishes just before the final pull on the handles. This point was found by stepping back through the forces from the last local minimum (stopping peak) until the force was greater than zero. These points are shown in Figure 4-1.

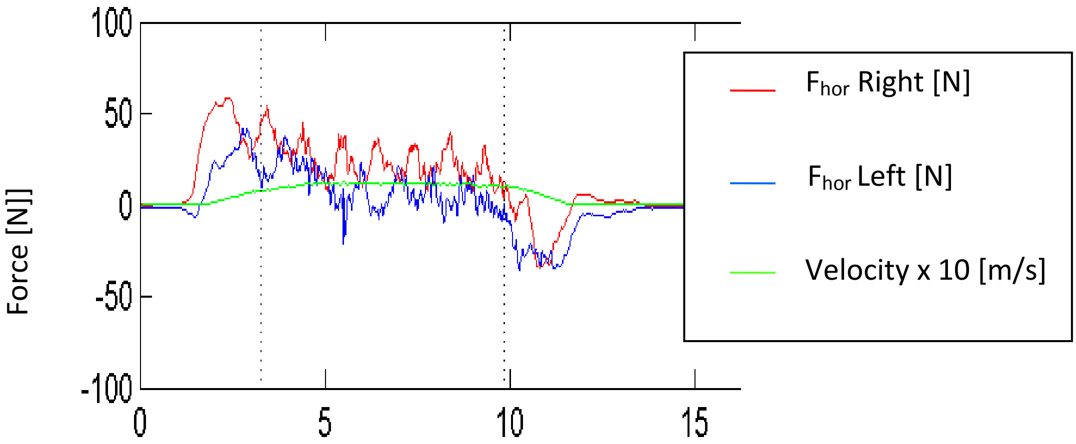


Figure 4-1: Example plot of left and right horizontal forces for an attendant –propelled run along with the velocity.

4.2 Provided Capabilities on Crossfalls

In order for them to overcome the required capabilities detailed in Appendix 3, they would need to:

1. Produce the peak force necessary when starting the wheelchair.
2. Produce the work necessary to move the wheelchair the required distance during the going phase.
3. Produce the peak force necessary when stopping the wheelchair.

It was further thought that when traversing a crossfall there would need to be a difference in force applied to the wheel which was 'upslope' compared to the one which was 'downslope'. The approach is similar to that taken for the self-propulsion experiments (see Figure 3-4). This leads to a number of hypotheses, which are detailed in Section 4.3.

4.3 Hypotheses

There are half the number of hypotheses in this chapter as there was in Chapter 2 due to the fact the occupant mass was fixed, and therefore is not considered as an independent variable in the multiple regression analysis.

The first set refer to the starting phase (Section 4.3.1); the second set to the going phase (Section 4.3.2); and the third set to the stopping phase (Section 4.3.3). If a significant difference is found for any of the hypotheses, it is possible that the difference was the result of a change to the forces, and in turn the work, done on the upslope and/or the downslope side of the wheelchair. For this reason, when significant results occur, they will be followed by a formal test on the upslope and downslope parameter.

4.3.1 Starting Phase

Hypothesis 31 and Hypothesis 32 test the effect of crossfall gradient on the provided capabilities of the APWS for the Starting phase.

Hypothesis 31.

H0: There will be no change in the Provided Starting Force regardless of crossfall gradient.

H1: There will be a significant proportional linear relationship between the provided starting force and crossfall gradient.

Hypothesis 32.

H0: There will be no change in the provided starting force difference regardless of crossfall gradient.

H1: There will be a significant proportional linear relationship between the provided starting force difference and crossfall gradient.

4.3.2 Going Phase

Hypothesis 33 and Hypothesis 34 test the effect of crossfall gradient on the provided capabilities of the APWS for the going phase.

Hypothesis 33.

H0: There will be no change in the provided going work regardless of crossfall gradient.

H1: There will be a significant proportional linear relationship between the provided going work and crossfall gradient.

Hypothesis 34.

H0: There will be no change in the provided going work difference regardless of crossfall gradient.

H1: There will be a significant proportional linear relationship between the provided going work difference and crossfall gradient.

4.3.3 Stopping Phase

Hypothesis 35 and Hypothesis 36 test the effect of crossfall gradient on the provided capabilities of the APWS for the stopping phase.

Hypothesis 35.

H0: There will be no change in the provided stopping force regardless of crossfall gradient.

H1: There will be a significant proportional linear relationship between the provided stopping force and crossfall gradient.

Hypothesis 36.

H0: There will be no change in the provided stopping force difference regardless of crossfall gradient.

H1: There will be a significant proportional linear relationship between the provided stopping force difference and crossfall gradient.

4.4 Ethics

The experiments were approved by the Ethics Committee at University College London. People were eligible for recruitment if they were over the age of 60, had no history of back pain (last 6 months) and who felt comfortable pushing a wheelchair with an occupant mass of approximately 75 kg along a 10 m pavement, with a crossfall gradient. Participants were recruited via an email that was sent out to the whole of UCL.

4.5 Equipment

The equipment used in these experiments will now be described, beginning the wheelchair used and its bespoke instrumentation in section 4.5.1. Details of how the push force were calculated from the force transducers are given in section 4.5.2.

The layout for these experiments and video recording system were identical to those for the self-propelled experiments (see Section 3.7. and Section 3.6.1 respectively).

4.5.1 The Wheelchair & Instrumentation

The wheelchair used in this experiment was a standard issue NHS attendant-propelled wheelchair, the 9L (see Figure 4-2). The wheelchair has a wheelbase (distance between rear axle and caster axle) of 36 cm when the casters are trailing back (as shown in Figure 4-2). The rear wheel track (distance between the rear wheels) is 50 cm. The total mass of the wheelchair system including the dummy and the equipment was 104 kg.

Figure 4-2: Attendant-propulsion wheelchair experiment system on left with detail of the force transducer (top right) and the rotary encoder used to measure the velocity (bottom right).

The wheelchair was instrumented so that the handle forces and also the rear wheel speeds could be recorded. The handle forces were recorded by attaching two 6-axis force transducers in line with the rubber grips of the handles. The force transducers used are commercially available and produced by AMTI (model MC3A-6-250). The force signals were amplified using amplification boxes again from AMTI (model MSA-6). The force data from the direction of travel (F_x) from both handles was recorded using a datalogger (Measurement Computing, model USB-5201). Both rear wheels had a rotary encoder to detect the velocity. The encoder consisted of a teethed gear which rotated with the rotating rear wheel. The encoder outputs 500 pulses per revolution and this signal was processed using custom circuitry. The resulting voltage was output to the same datalogger as the one used to collect the force data. The accompanying datalogger software (TracerDAQ) was used to record the data files, which was run on a laptop secured to the wheelchair. All data was recorded at a sampling frequency of 100 Hz.

The data-logger files contain the right and left wheel velocities as well as the right and left F_y output from the force transducers. The F_y output is the main component of the horizontal force (F_{hor}).

4.5.2 Calculating push Force

The handles of the wheelchair are naturally at a 21 degree angle from the horizontal. This is illustrated in Figure 4-3,which shows F_y, F_z, F_{ver} and F_{hor} . F_y is the force measured along the y-axis of the force transducer,F_z is the force measured along the z axis of the force transducer, F_{ver} is the vertical force and F_{hor} is the horizontal force. F_{ver} was calculated using Equation 14 and F_{hor} was calculated using Equation 15

$$F_{ver} = F_y sin\theta + F_z cos\theta$$

Equation 14: Equation to calculate the vertical force component from the readings from the y-axis (F_y) and x-axis (F_x) force transducers. Θ is the angle of inclination of the handle to the horizontal.

$$F_{hor} = F_y cos\theta + F_z sin\theta$$

Equation 15: Equation to calculate the horizontal force component from the readings from the y-axis (F_y) and x-axis (F_x) force transducers. Θ is the angle of inclination of the handle to the horizontal.

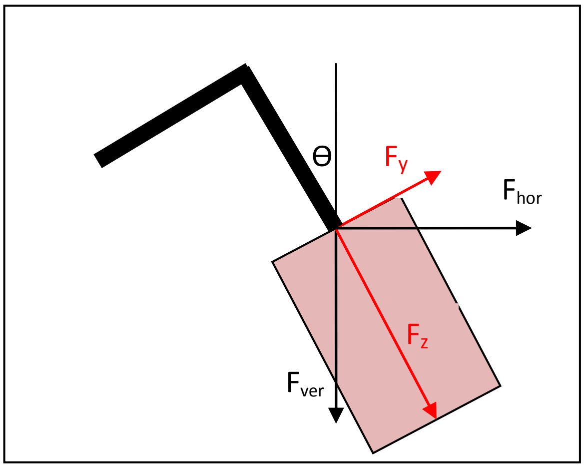


Figure 4-3: Schematic representation of recorded and calculated forces of the handle. F_y and F_z are the components of the force in the y and z axis respectively. F_{ver} is the vertical force and F_{hor} the horizontal (push) force.

4.6 Protocol

The protocol for the maximum voluntary push force test and the crossfall experiments will now be described.

4.6.1 Maximum Voluntary Push Test (MVPF)

A maximum voluntary push test was completed by each participant to capture the maximum amount of force the participant was capable of applying to the push rim when the wheelchair is restrained from moving.

For the Maximum Voluntary Push Test (MVPF) the wheelchair was placed up against the parapet at the side of the PAMELA platform and the brakes were applied to the wheelchair to prevent it from moving. The wheelchair was placed up against the parapet of the PAMELA platform and the brakes applied to prevent it from moving. As the handles are higher than the centre of mass of the system a box was placed between the wheelchair and the parapet to prevent the wheelchair rotating as the attendant pushed it.

They were then given the following verbal instructions:

"When I tell you to 'GO' I want you to push the handles of the wheelchair as hard as you can 3 times. The wheelchair should not move. Please push for a count of 3 seconds with a rest of 5 seconds between each push. I will count and time you. When you have finished the 3 second push please remove your hands from the wheelchair. Do you have any questions?" PAUSE "GO."

The force data was collected though a laptop connected to the amplification boxes of the force transducers and was collected using NetForce software.

4.6.2 Crossfall Experiments

The participant was asked to position the wheelchair in the correct starting position. For this to be the case the wheelchair needed to be directly behind the start line, the casters and the wheelchair had to be parallel to the intended direction of travel, the casters must be trailing backwards and the dashed red line had to be mid-distance between the two casters. On occasion help was given to ensure the casters were orientated in the right direction and the wheelchair was in the correct starting position.

Participants were asked to stand behind the wheelchair with their hands not touching the handles before each trial. This was done for easy identification of the start of the run.

Participants were then given the following verbal instructions before completing each for the 3 test conditions:

“When I tell you to ‘GO’ I want you to push the wheelchair in a straight line by attempting to follow the dashed red line on the floor. Push at a comfortable speed, as if you were pushing on a path. Keep pushing until you pass the stop line. Then stop as quickly as you are able and do not turn the wheelchair. Please remove your hands form the handles when you have finished. Do you have any questions?” PAUSE “ GO.”

The data for the force transducer and the velocity encoders were recorded between runs. The participant was then asked to position themselves for the next run, with help given as required.

4.7 Data Analysis Methods

4.7.1 Maximum Voluntary Push Test (MVPF) data reduction

The data files from NetForce were imported into Matlab, where the Push Force was calculated using Equation 15. The peak value of the push force was then found using the Matlab’s ‘max’ function.

4.7.2 Data Analysis Methods: provided capabilities

A custom Matlab script was written to analyse the data files. This read in the data files from the Datalogger and also the NetForce files. Each file contains the Fy component of the force from the right and left handles and these were used to synchronise the data. This was done by finding the first local maximum of Fy in the datalogger file and finding the same maximum in the NetForce file. The data for the NetForce file was then shifted so that the 2 maxima coincided. Both sources of data were collected at 100 Hz.

The velocity data was filtered using a 4^{th} order Butterworth filter with a cut-off frequency of 5Hz. This relatively low cut-off frequency was needed to eliminate the high frequency vibrations which were picked up by the rotary encoders.

4.7.3 Statistical Analysis

The statistical tests carried out were identical to those for the self-propelled experiments (see 1.9.4, except that the resulting linear regression moderAll statistical analysis was carried out using PASW Statistics 18, Release Version 18.0.0 (SPSS Inc., 2009, Chicago, IL, www.spss.com).

Where data sets were normally distributed multiple linear regression analyses were run. Each parameter (e.g. Provided Start Force) was defined as the independent variable. Crossfall gradient (C) was defined as the dependant variable (also termed regressor). This analysis produces a prediction equation based on the regression coefficients (β) and the independent variables (Tabachnick 2001).

$$\mathrm{P} = A + \beta_c C$$

Equation 16: prediction equation for parameter (P) when dependent variables are crossfall gradient (C), and participant weight (W). A is the constant term.

5 Results: Self-Propulsion

5.1 Participants

Fourteen participants took part in the study. Twelve were able-bodied and two were regular wheelchair users; one of whom had had polio as a child and the second had had an accident resulting in a Spinal Cord Injury (see Table 3-1). There were 12 females and 2 males. The average weight of the participants was 69.08 kg with a standard deviation of 14.86 kg. Eleven out of the fourteen participants were right handed.

Table 5-1: Participant details for self-propulsion experiments.

Participant Number	Age [Years]	Weight [kg]	Right or Left Handed	Male (M) or Female (F)	Wheelchair user?
1	28	59.90	Right	F	No
2	28	54.30	Right	M	No
3	34	64.00	Left	M	No
4	28	100.75	Right	M	No
5	35	71.70	Right	M	No
6	53	66.05	Right	M	No
7	53	49.20	Right	M	Yes
8	28	88.40	Right	M	No
9	49	84.85	Right	M	Yes
10	27	51.75	Right	M	No
11	36	65.70	Right	M	No
12	24	71.10	Left	M	No
13	29	79.40	Right	M	No
14	33	60.00	Left	F	No

5.2 Maximum Voluntary Push Force

As explained in Section 2.5, the Maximum Voluntary Push Force (MVPF) is an indicator of the maximum Provided Capability of the SPWS. This is measured using a MVPF test as described in Section 1.8. An example plot of the MVPF test is given in Figure 5-1 and the results for the MVPF are summarised in, with the two regular wheelchair users highlighted in green.

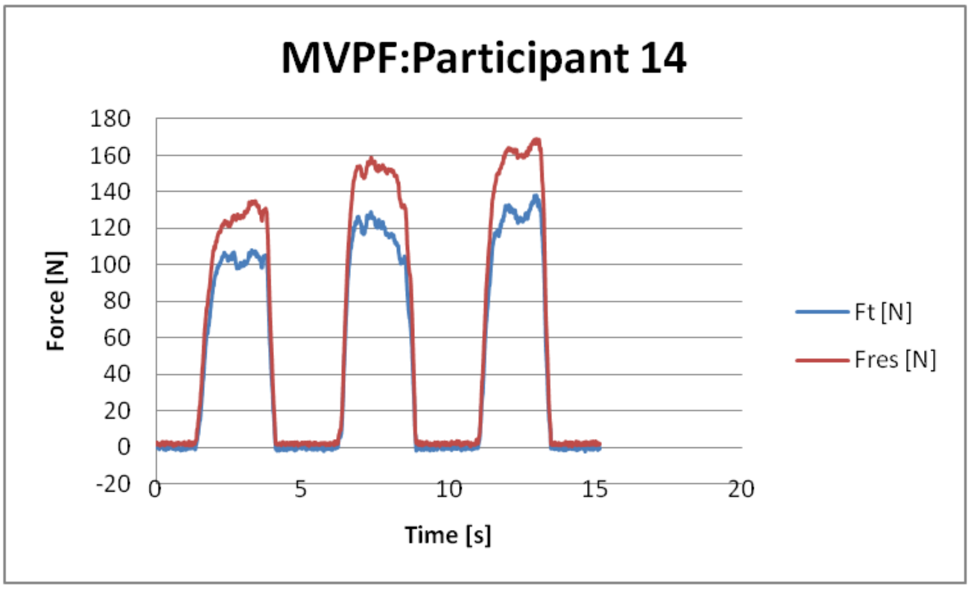


Figure 5-1: Example MVPT plot for self-propulsion showing the resultant force (Fres) and tangential force (Ft)

The maximum tangential force (F_t) applied to the handrim varied from 88.26 N to 244.14 N with a mean of 162.41 N (see Table 3-2). There was no apparent relationship between peoples' mass and their ability to produce force as measured by the MVPF, as can be seen in Figure 5-2.

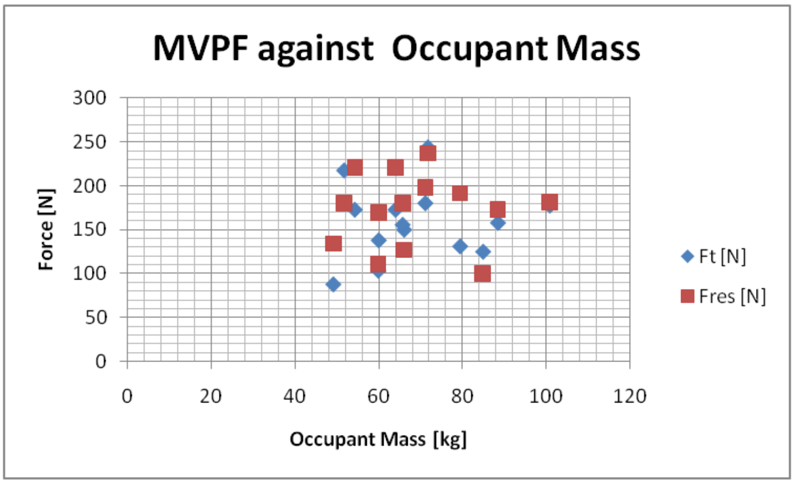


Figure 5-2: Maximum tangential and resultant forces from MVPF plotted against occupant mass

Table 5-2: Summary of results of voluntary maximum push.

Participant	System Weight [N]	Ft [N]	Ftot [N]
1	758.31	103.82	110.67
2	703.38	173.17	220.71
3	798.53	173.17	220.71
4	1159.05	177.87	181.11
5	874.07	244.14	237.14
6	818.64	150.31	127.06
7	653.35	88.26	133.9
8	1037.90	158.18	172.71
9	1003.07	125.34	99.79
10	678.36	217.83	180.45
11	815.21	155.97	180.52
12	868.19	180.56	197.95
13	949.61	131.48	191.34
14	759.29	138.4	169.37
Mean	848.35	158.46	173.10
Standard deviation	145.75	41.49	41.79

It would appear based on these results, compared with the Required Capabilites of the task (see results of Appendix 3) that all occupants are able to impart sufficient force into the wheelchair to produce the Provided Capbility to enable it to begin to move and to continue to move.

5.3 Deviation from a straight line

The participants were asked to travel in a straight line along each of the footways. When this was not achieved, the deviation was observed during the experiments. The values noted during the experiments were checked by reviewing the experiment videos as necessary (see Section 1.9.2). It is important that the occupant be able to complete the task when the downslope side of the wheelchair is their non-dominant hand, as well as when it is their dominant hand.

The deviation from the straight line is, in effect, the output from the interaction between the provided capabilities of the SPWS and the required capabilities of the environment. Therefore if a straight line was achieved the provided capabilities were greater than the required capabilities.

On the 0 % condition all participants successfully managed to go in a straight line, with no noticeable deviation. The only person to truly struggle on the 2.5 % and 4 % conditions was Participant 1. She had a maximum deviation of 1.2 m, on the 2.5% crossfall and deviated the full width of the footway on all six trials on the 4% crossfall (see Table 3-3). She commented that only the 0% condition was “easy”, the 2.5% ”a little tricky” and that she felt ”out of control” on the 4% crossfall even though she was “constantly adjusting how [she] pushed”.

Of the other participants 4 deviated from the straight line on the 2.5% and 4% crossfalls (see Table 5-3). However, apart from Participant 1, nobody deviated by more than 0.4 m. All bar participants 1 and 3 managed to complete the task without a noticeable deviation for at least 1 trial travelling in both directions.

Table 5-3: Table of observed straight line deviations for participants 1,3,6,7, and 8 for each run from 1-6. Red text: highlighting the relatively large deviations made by participant 1 relative to all other participants.*participant stopped twice

P#	2.5%						4%					
	R1 [m]	R2 [m]	R3 [m]	R4 [m]	R5 [m]	R6 [m]	R1 [m]	R2 [m]	R3 [m]	R4 [m]	R5 [m]	R6 [m]
1	0.4*	0.4	1.2	0.8	0.8	0.8	1.2	1.2	1.2	1.2	1.2	1.2
3	0.4	0.4	0.4	0.4	0.4	0.4	0.4	0	0.4	0.4	0.4	0.4
6	0	0	0	0	0	0.4	0.4	0	0.4	0	0.4	0
7	0.4	0	0.4	0	0	0	0	0	0	0	0	0
8	0	0	0	0	0	0	0	0.4	0	0.4	0	0

As Participants 1 and 3 did not accomplish the task when their non-dominant hand was on the downslope side (see Table 5-3), it must be concluded they were unable to do so with the wheelchair used in this study. Therefore, the provided capabilities of the SPWS are insufficient to complete the task. All other participants were successful, so it can be concluded that in these cases the SPWS had the provided capabilities needed to achieve the task.

The next section will investigate the individual provided capabilities used to travel along a footway, beginning with provided going work and provided going work difference.

5.4 Starting & stopping

In general the starting and stopping forces were larger and longer than the forces used in the going phase when on the flat. However, on the downslope side this distinction became less clear for the starting force. There are rarely any brakes in the going phase to compare to that found in the stopping phase, this is not the case when a crossfall is present, where brakes can frequently occur (see Figure 5-3D).

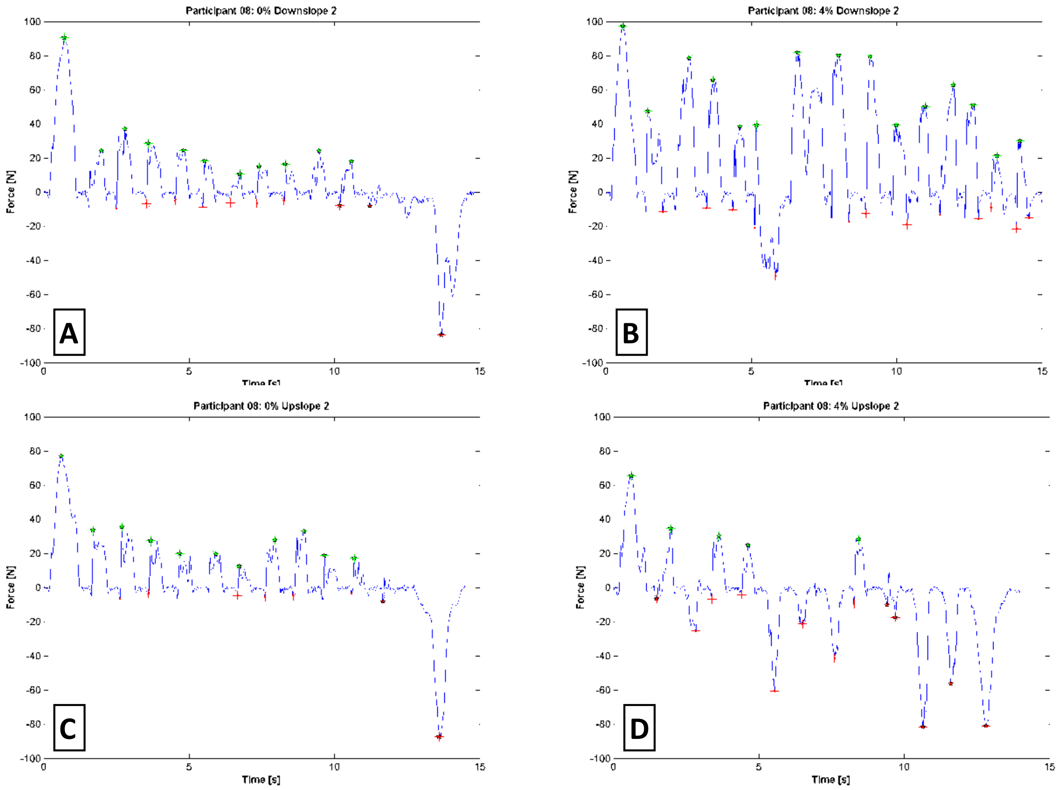


Figure 5-3: Illustration of how the start pushes force is distinct from the going pushes when the wheelchair is on the 0% crossfall (A=Downslope, C=Upslope). In B the pushes are similar in size and duration to those in A. In D the brakes in the Going phase are similar in magnitude and duration to the stopping force.

There was a large degree of variance in the magnitude and the duration of starting and stopping forces used by occupants, this resulted in the data not being normally distributed and Hypothesis 1 to Hypothesis 12 are tested using non-parametric tests as detailed in Section 3.9.4. These hypotheses, which test the effect of crossfall gradient on Starting and stopping forces, will now be tested for the downslope (Section 5.4.1 and the upslope sides (Section 5.4.2) of the wheelchair.

5.4.1 Downslope Starting and stopping

The magnitude of the average starting force did not change significantly across the 3 crossfall gradients ($X^2(2) = 2.714$, $p<.257$). The push time also did not change significantly ($X^2(2) = .764$, $p<.683$). See Table 5-4 for median values.

The peak force used to stop the wheelchair did not change significantly with crossfall gradient ($X^2(2) = 3.857$, $p<.145$). A shorter stopping force was found to be applied as crossfall increased ($X^2(2) = 7.00$, $p<.030$). However, post-hoc tests did not find any significant differences between the three groups.

Independent Variable	0%	2.5%	4%
Downslope Start Ft [N]	72.08	65.84	69.78
Downslope Start Push Time [s]	.995	.960	.930
Downslope Stop Ft [N]	-42.37	-38.31	-40.02
Downslope Stop Push Time [s]	1.94	1.92	1.65

Table 5-4: Median downslope starting and stopping forces for each crossfall condition

5.4.2 Upslope Starting and stopping

The average peak starting force required decreased as crossfall gradient increased on the upslope side ($X^2(2) = 8.714$, $p<.013$). When the three crossfall gradients were compared, post-hoc, with the use of a Wilcoxon Signed-Rank Test, it was found there was a significant decrease from 0% to both the other conditions ($Z = -2.919$, $p=.004$) and ($Z = -2542$, $p=.011$) respectively for the decrease to 2.5% and 4%). The push time for the Starting push also decreased with crossfall gradient ($X^2(2) = 7.964$, $p<.017$) and again this was significant from 0% to 2.5% ($Z = -2.797$, $p=.005$) and from 0% to 4% ($Z = -2.450$, $p=.014$), but not from 2.5% to 4% ($Z = -.785$, $p=.433$).

The force applied to stop the wheelchair did not change significantly in magnitude ($X^2(2) = .429$, $p<.807$) but did get significantly longer as crossfall increased($X^2(2) = 6.67$, $p<.036$). Post-hoc tests showed there was a significant difference between 0% and 4% ($Z = -2.919$, $p=.004$) (see Table 5-5 for median values).

Table 5-5: Median downslope starting and stopping forces for each crossfall condition. The following key is used: ^ indicates significant differences between 0 % and 2.5 % crossfalls, $ indicates significant differences between 0 % and 4 % crossfalls.

Independent Variable	0%	2.5%	4%
Upslope Start Ft [N]	63.49$^	52.42^	50.74$
Upslope Start Push Time [s]	.950	.840	.895
Upslope Stop Ft [N]	-56.14	-62.12	-61.74
Upslope Stop Push Time [s]	1.97$	2.50	2.57$

In contrast to the downslope contact parameters for start-up and stopping (which remained consistent regardless of crossfall gradient) upslope start-up forces changed in both magnitude and time applied as crossfall increased. The time of the stopping force also increased on the upslope side. Therefore, it would seem people preferred to reduce the amount of force done at start-up on the upslope side, than increase force on the downslope side. Also, people preferred to apply the stopping force over a longer time on the upslope side as crossfall increased. The provided capabilities used to keep the wheelchair in motion will now be reported.

5.5 Provided Capabilities of the Going Phase

The provided capabilities to keep a wheelchair moving in a straight line along a footway can be expressed in terms of the amount of work produced in order to (in most cases) successfully propel the wheelchair along a crossfall. Work can be thought of as the transfer of energy from the occupant to the wheelchair, which results in the SPWS moving along the footway.

Four hypotheses were proposed in regarding the effect of crossfall gradient and occupant mass on the provided capabilities during the going phase: the provided going work (C_{wk_sum}) and the provided going work difference (C_{wk_diff}). The hypotheses were tested as described using a Multiple Regression Model (see section 3.9.4 for details).

The results of the regression models for C_{wk_diff} and C_{wk_sum} using crossfall gradient and occupant mass as regressors are given in Table 3-6. For C_{wk_diff} two models were considered, the first contained both crossfall and occupant mass as regressor terms (row 2 Table 3-6), and the second contained only crossfall (row 3 Table 5-6). The format of these tables will be the same for all parameters reported in this thesis with the Dependant variable in the left-hand column, the next column (to the right) will contain the R^2 coefficient, which tells one how much of the variance the model is able to explain. The subsequent column gives the significance level of the R^2 value. The columns to the right of the table report the β-coefficients of the independent variables along with their significance level (as measured using a t-test).

Table 5-6: Regression model summary for the provided going work (Cwk_sum) and the difference of work (Cwk_diff).

Dependant Variables	Model		Coefficients					
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P	Occ. Mass [kg]	P
C_{wk_diff}	.865 (.863)	<.0001	-51.119	<.0001	20.174	<.0001	.684	<.0001
C_{wk_diff}	.797 (.795)	<.0001	-3.912	<.0001	20.229	<.0001	N/A	N/A
C_{wk_sum}	.063 (.047)	.02	71.791	<.0001	-2.662	.093	.415	.023

5.5.1 Provided Going Work

The regression model for provided going work was a very poor degree of fit (R^2=.063, R_{adj}^2=.047), and although the relationship is significant (F(2,123)=4.04, p=.02), meaning statistically the model has predictive ability, it is only capable of modelling approximately 5% of the variance recorded in C_{wk_sum} and so is not a generally useful model. In fact the mean values for each condition are very similar: 104.09 Nm, 84.12 Nm and 96.30 Nm respectively for 0%, 2.5% and 4%, with no particular trend visible (see Figure 5-4).

Analysis of the independent variables, with the use of a t-test, proved there was no effect of crossfall on C_{wk_sum} (p=.093). occupant mass was found to have a positive correlation with the provided going work necessary, with an additional 0.415 Nm provided for each kg

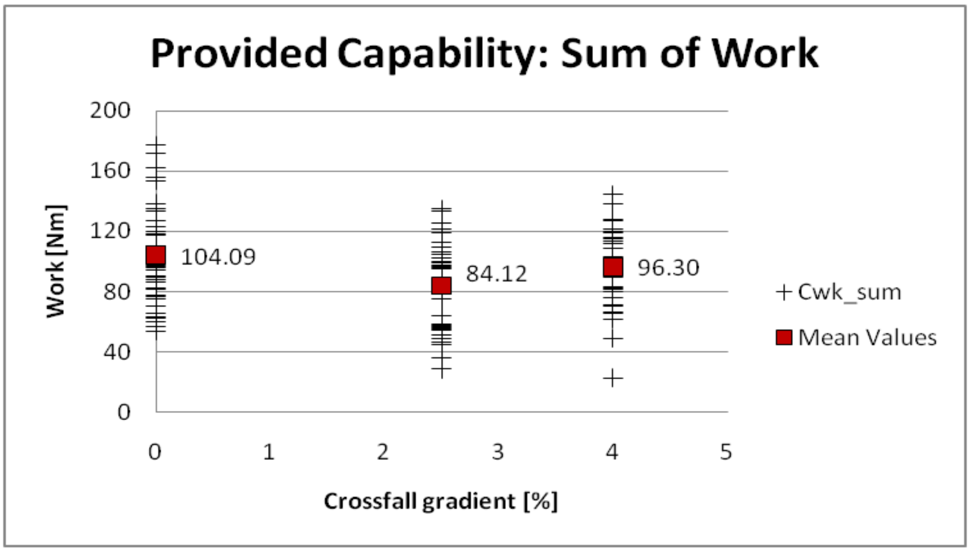


Figure 5-5: The capability to produce the sum of work (C_{wk_sum}) done on the upslope and downslope runs plotted against crossfall gradient, with mean values for each condition displayed in red with the accompanying value.

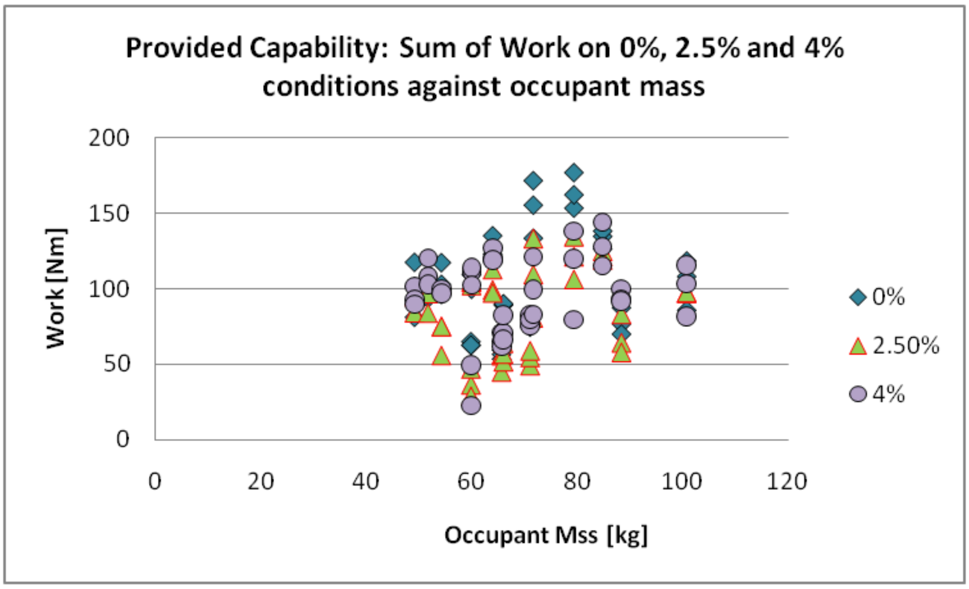


Figure 5-6: Sum of upslope and downslope work against occupant mass.

increase in mass (p=.023). This general trend is visible in Figure 5-5; though it is clear there is a great deal of variation within the data.

In conclusion, the regression model is unable to predict a useful amount of variance and Hypothesis 1 for the effect of crossfall gradient on the C_{wk_sum} can be accepted: there is no significant effect of crossfall gradient on the C_{wk_sum}. However, the null hypothesis for the effect of occupant mass can be rejected: there is a small but significant increase in C_{wk_sum} as occupant mass increases.

5.5.2 Provided Going Work Difference

The results of the multiple linear regression of the effect of crossfall gradient on the provided going work difference (C_{wk_diff}) are shown in Table 3-6. The model has a reasonable degree of fit (R^2=.797, R_{adj}^2=.795), and was significant at explaining the variation in the data (F(2,123)=478.194, p<.0001). There was a small negative constant term of -3.912 Nm (p<.0001) whereas there was a large positive coefficient for crossfall gradient of 20.229 Nm (p<.0001). The individual trial data is plotted in Figure 5-6 along with the regression line.

When occupant mass was added as a regressor term the amount of variation the model was able to predict increased, as would be expected (F(2,123)=388.914, p<.0001) and the regression model was a good fit (R^2=.865, R_{adj}^2=.863). The results of the model are given in Table 5-6.

When the individual variables are looked at with the aid of a t-test both crossfall and occupant mass are significant. Crossfall was positively correlated to C_{wk_diff}, with a correlation coefficient of 20.174 Nm (p<.0001) (see Table 3-6. Occupant mass was also positively correlated to C_{wk_diff} (p<.0001) but the actual influence was low, with an increase of 0.684 Nm for every kg increase in mass. The equation for the model is given in Equation 6.17.

$$C_{wk\_diff} = -51.116 + 20.174(C) + 0.684(M)$$

Equation 6.17: Regression equation for capability required to apply differing force to upslope and downslope handrims. C is the crossfall gradient as a percentage, M is the mass of the occupant in kilograms.

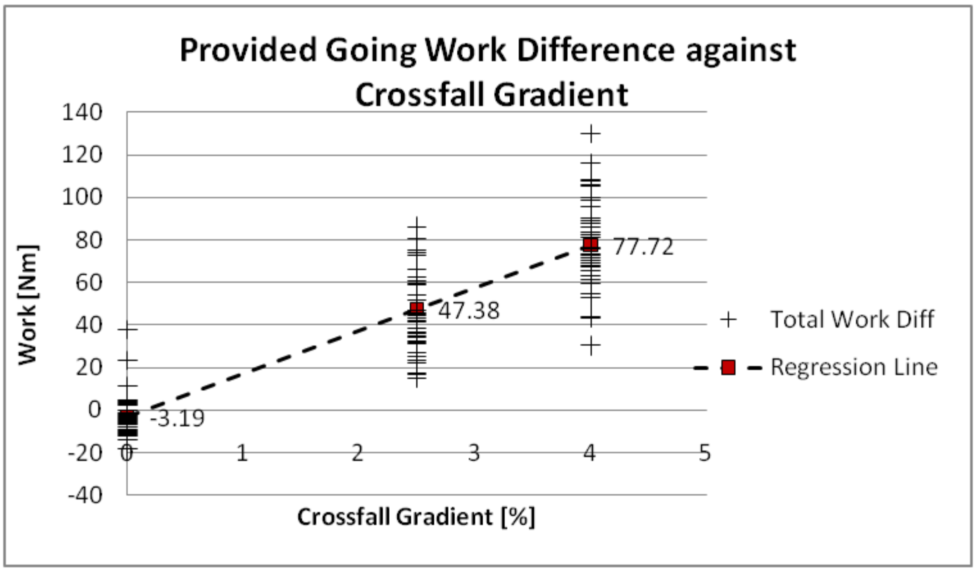


Figure 5-7: Difference of Work between downslope and upslope runs against crossfall gradient, along with the regression line.

Although the model has statistical predictability, as can be seen in Equation 6.17 there is a rather large negative constant term of -51.116 Nm (p<.0001). This may be a little counterintuitive as it might be thought that there was a difference of over 50 Nm between the amounts of work done on the downslope side compared to the upslope side on the 0% condition. However, when the effect of adding occupant mass is taken into account, the constant term makes more intuitive sense. This is illustrated in Figure 5-7 where the range of occupant masses is plotted in 10 kg steps using Equation 6.17 .

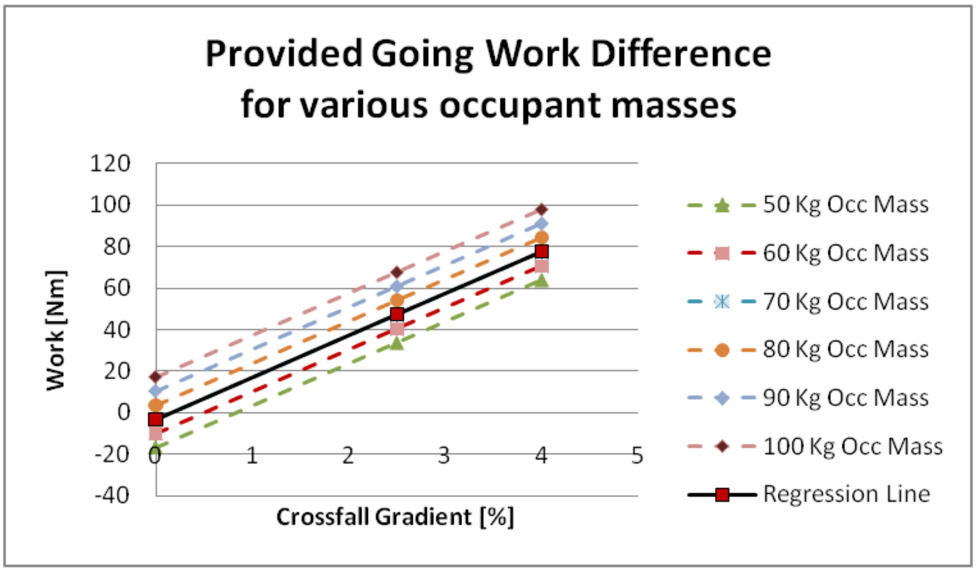


Figure 5-9: Calculated values of the provided going work difference (C_{wk_diff}) on the upslope and downslope runs from the results of the regression models plotted against crossfall gradient. The first 6 series are calculated using crossfall and weight as regressors and the final series ('Regression Line') is the results of the regression model when only crossfall is a regressor. See Table 3 for details of the models.

In Figure 5-7 the regression line for the simpler model, with only crossfall as an independent variable, is also shown (as solid line) and lies in the middle of the range of occupant masses. There is a range of C_{wk_diff} from -20Nm to +20Nm on the 0% condition, which corresponds to the range seen in the original data (see Figure 5-6). However, the model predicts that lighter people would produce a negative amount of C_{wk_diff} and heavier people a positive amount of C_{wk_diff} . It is far more likely that any offset on the 0% condition would be due to right or left

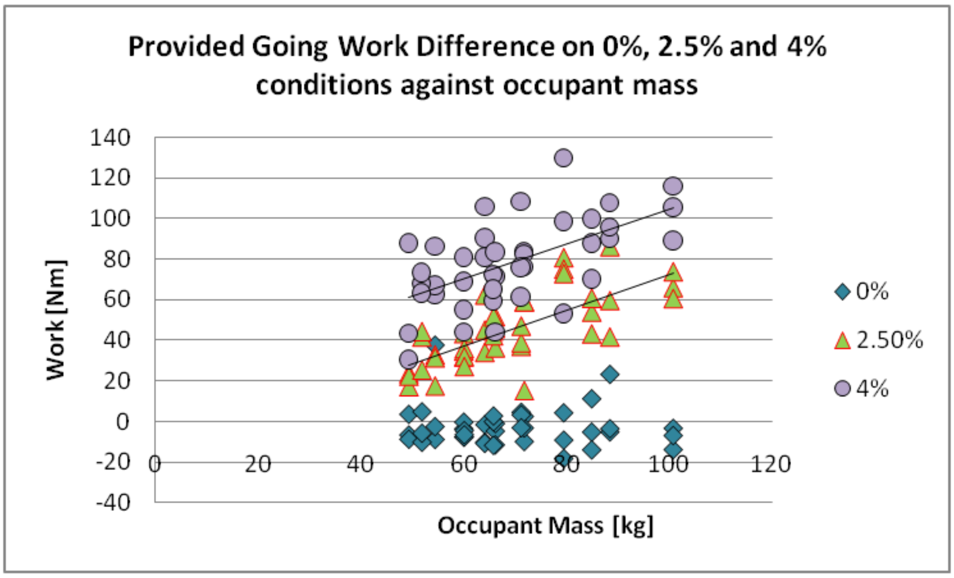


Figure 5-8: Individual measured difference of Provided Going Work Difference between downslope and upslope runs against occupant mass, with trendines shown for the 2.5% and 4% crossfalls.

hand dominance than a person's weight. This is not to say that there is not an effect of occupant mass (see Figure 5-8).

To sum up, there is a significant effect of crossfall and occupant mass on C_{wk_dif} and therefore the null hypotheses that C_{wk_dif} is unaffected by these variables, can be rejected. The amount of C_{wk_dif} increases approximately 20 Nm with each percentage increase in crossfall gradient. However, it is less clear how much C_{wk_dif} increases with occupant mass, though it is clear there is a general trend of increasing C_{wk_dif} with increasing crossfall gradient. Therefore, we now know there has been a significant increase in provided going work difference as occupant mass and crossfall increase, the source of this increase will be investigated, by investigating the amount of positive work done on the downslope side (Section 5.5.2.1) and the amount of positive (Section 5.5.2.2) and negative work (Section 5.5.2.3) done on the upslope side. The amount of negative work on the downslope side is not investigated as there was virtually no negative force applied to the handrim during the going phase of each run on the downslope side.

5.5.2.1 Downslope Positive Work

The amount of Downslope Positive Work provided by the SPWS to the task for each run was calculated by summing the positive work done during a run. Work was calculated as described in section 2.7.2. The average value for each person (of the 3 runs) is shown in Table 3-7.

There is a general trend of increasing Downslope Positive Work as crossfall increases, which is visible both in Table 3-7 and Figure 5-9. All participants used a much larger amount of Downslope Positive Work on the 4% crossfall, compared with the 0% and 2.5% crossfalls. However, participants 2, 5, 6, 7 and 12 all used similar values on the 0% and 2.5% crossfall, and these cases are highlighted in blue in Table 5-7.

Table 5-7: Mean values of Downslope Positive Work for each participant and each crossfall gradient, showing a general trend of increasing work on the 4% crossfall compared with the other 2 gradients. The cells highlighted in blue show the cases where the participant applied similar values of work on the 0% and 2.5% crossfalls.

	Downslope Positive Work [Nm]		
Participant	Mean 0%	Mean 2.5%	Mean 4%
1	32.64	44.28	66.99
2	60.25	52.12	81.56
3	61.61	74.32	107.05
4	54.67	91.60	116.65
5	79.81	86.50	106.05
6	57.40	54.57	84.44
7	56.24	60.76	75.34
8	47.06	70.73	97.10
9	78.29	104.79	131.70
10	51.58	67.43	95.52
11	33.36	56.52	78.36
12	46.18	51.43	78.66
13	81.39	107.05	119.44
14	57.65	73.22	86.02
Mean	57.01	71.09	94.63

The observed tendency of increasing downslope positive work with increasing crossfall gradient was statistically tested using a multiple linear regression with crossfall gradient and occupant mass as regressors.

The resulting model from this analysis was a reasonably good fit (R_{adj}^2=.527) and the relationship was significant (F(2,121)=69.464, p<.0001). Both crossfall and occupant mass were found to have a significant (p<.0001) relationship with downslope positive work provided by the users (see Table 3-8).

The constant term was not significant (p=0.121). The fact the constant term was not significant could be due to the similar values of work on the 0% and 2.5% for 5 of the participants (see Table 3-8). However, it is more probably due to the amount of variability between people which still exists even when their mass is taken into consideration. It should be noted that the use of predominantly non-wheelchair users means that the range of downslope positive work are more homogenous than if a full range of wheelchair users were used, each with a different mobility impairment

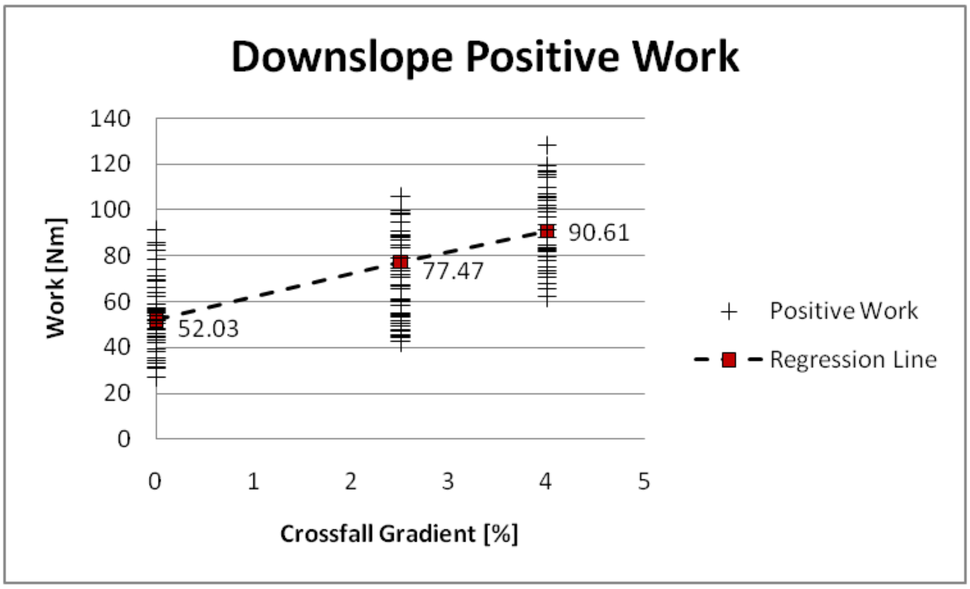


Figure 5-10: Individual runs of Downslope Positive Work, along with the regression line using coefficients from table 6 where only crossfall is used as a regressor term.

Although the model is unable to estimate the constant term with a suitable level of significance, the coefficients representing the slope of the line are significant; there was an additional 8.76 Nm needed with each percentage increase in crossfall, and .59 Nm required with each kg increase in occupant mass. Therefore, propelling on a crossfall of 2.5% would require additional 21.9 Nm of downslope work, and an extra 35 Nm would be needed on the 4% according to this model. By way of comparison the increase in downslope positive work with each percentage increase in crossfall gradient is approximately equivalent to a 37 kg increase in occupant mass using the coefficients from this model Table 3-8.

Table 5-8: Multiple Regression Analysis for Downslope Positive Work on downslope runs, showing that the model is capable of explaining over 50% of the variance seen in the data, and that crossfall and occupant mass both have significant (P<.0001) positive coefficients, while the constant term is not significant.

Dependant Variables	Model		Coefficients					
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P	Occ. Mass [kg]	P
Positive Work	.534 (.527)	<.0001	11.321	.121	8.759	<.0001	.588	<.0001

It can be concluded that the amount of downslope positive work significantly increased as crossfall gradient and occupant mass increased. The increase in downslope positive work was estimated to be 8.76 Nm per percentage increase. This falls short of the downslope positive work difference calculated in Section 3.5.2 (pg. 87) used to continue along a straight line, which was estimated to be 20 Nm.

As stated in Section 2.5.2 the second way in which the downslope positive work difference could have been achieved would have been to change the amount of work on the upslope side. This will now be investigated.

5.5.2.2 Upslope Positive Work

The Upslope Positive Work was calculated by summing the amount of work done on the upslope side. The work was calculated (as described in Section 2.7.2) for each run and the average of the 3 runs calculated. These average values are shown in Table 3-9 for each participant and each crossfall gradient.

Table 5-9: Mean values of Upslope Positive Work for each participant and each crossfall gradient, showing the amount of work, decreased when the crossfall increased from 0%. However, some participants used similar amounts of work on the 2.5% and 4% crossfalls; these cases are highlighted in blue.

Participant	Mean 0%	Mean 2.5%	Mean 4%
1	34.81	23.86	14.57
2	50.33	33.58	27.91
3	67.62	30.60	23.20
4	61.27	29.20	33.19
5	81.44	38.76	19.69
6	59.31	36.51	33.00
7	55.69	39.00	33.30
8	37.68	29.60	22.47
9	74.16	47.36	29.62
10	53.80	36.89	29.59
11	34.34	20.71	10.12
12	43.11	22.06	19.66
13	88.32	42.64	29.03
14	56.92	37.54	35.10
Mean	57.06	33.45	25.74

In Table 3-10 there is a trend of decreasing upslope positive work as crossfall increases. However, the decrease between 2.5% and 4% appears less severe than the decrease between 0% and 2.5%, and in some cases (highlighted in blue) the amount of work is actually quite similar.

The mean value of the 0% crossfall runs for the upslope positive work of 57.06 Nm is very similar to that found for the downslope positive work of (57.01 Nm), and the spread of data is likewise very similar, ranging from approximately 25 Nm – 85 Nm in both cases (see Figure 5-9 and Figure 5-10).

The relationship between upslope positive work and crossfall gradient, as well as occupant mass was tested using a multiple regression model. The model was a reasonably good fit

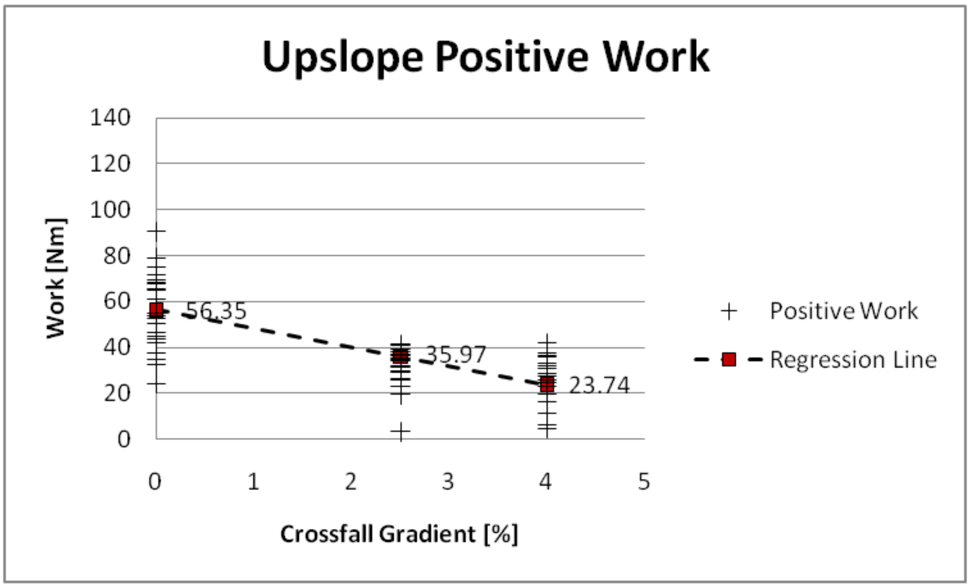


Figure 5-11: Individual runs of upslope positive work, along with the regression line using coefficients from Table 8 where only crossfall is used as a regressor term.

(R^2=.538, R_{adj}^2=.535) and the relationship was significant (F(2,121)=69.464, p<.0001). Crossfall was found to have a significant negative β-coefficient (p<.0001), however, occupant mass was not significant (p=.162) and for this reason the model was re-run without occupant mass as a regressor term. A summary of the original model is given in Table 3-10 along with a summary of the subsequent simpler model.

Table 5-10: Multiple Regression Analysis for Upslope Positive Work

Dependant Variables	Model		Coefficients					
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P	Occ. Mass [kg]	P
Upslope Positive Work	.538 (.535)	<.0001	48.707	<.0001	-8.162	<.0001	.111	.162
Upslope Positive Work	.538 (.535)	<.0001	56.354	<.0001	-8.153	<.0001		

The result of the regression model with just crossfall gradient as the regressor term was also a reasonable fit (R^2=.538, R_{adj}^2=.535) and the relationship was significant (F(2,121)=141.171, p<.0001). The results of this model are plotted in Figure 5-10. There was a significant negative β-coefficient for crossfall gradient of -8.153 Nm (p<.0001). This is similar in magnitude to that observed for the total positive work of 8.759 Nm (see Section 5.5.2.1). It can be concluded that there is a significant effect of crossfall, but not of occupant mass on upslope positive work.

The combination of a decrease in upslope positive work and an increase in downslope positive work result in a provided going work difference (C_{wk_diff}) of approximately 17 Nm, which is very close to the 20Nm estimated to be needed in Section 3.5.2 (pg. 87).

The third way of producing a difference in work between upslope and downslope sides is for the occupant to apply a negative force to the upslope side, resulting in negative work being done. This would, in effect, decrease the total work done on the upslope side. This upslope negative work caused by Brakes and Impacts will now be investigated (Section 3.5.2.3).

5.5.2.3 Upslope Negative Work

In Table 5-11 the median[17] values of the upslope negative work done for the 3 runs for each crossfall gradient and for each participant are shown in Table 3-11.

Table 5-11: Median Upslope Negative Work.

	Median Upslope Negative Work [Nm]		
Participant	0 %	2.5 %	4 %
1	-1.02	-23.56	-37.29
2	-1.83	-12.84	-11.12
3	-0.72	-2.55	-5.09
4	-5.02	-13.91	-37.55
5	-2.76	-6.95	-8.09
6	-11.43	-14.26	-35.03
7	-3.53	-3.63	-5.50
8	-0.93	-23.05	-20.57
9	-6.09	-12.59	-14.27
10	-3.94	-8.63	-7.86
11	-4.40	-17.60	-11.39
12	-4.76	-15.48	-13.82
13	-2.16	-20.36	-28.82
14	-0.50	-1.31	-5.76
Median (IQR)	3.15	-13.38	-12.61

The first thing to note is that for nearly all participants (bar participant 11) the amount of upslope negative work increased. However, for participants 3, 7, and 14 the amount of

[17] Median values as opposed to mean values are shown here as the data was found not to be normally distributed. As a non-parametric test was used to test the data, which in effect ranks the values it was thought the median would be more useful and meaningful a parameter to report.

upslope negative work done on the 2.5% and 4% crossfalls falls within the 90th percentile of values for the 0% crossfall of -8.01 Nm; and participants 5 & 10 are only slightly beyond this value on the 4% crossfall. These 5 participants are highlighted in green in Table 5-11.

The second detail worthy of note is that some participants had similar values for upslope negative work on the 2.5% and 4% crossfall (these are highlighted in blue in Table 5-11). Two of these used less upslope negative work on the 4% crossfall compared with the 2.5% crossfall. However, participants 1, 4 and 6 all had large increases in upslope negative work as the crossfall gradient increased.

Therefore it could be concluded that there are in fact 3 different types of coping strategy used by people with regards to upslope negative work (shown in Table 5-11): those who apply very little negative upslope work (the green rows), those who apply similar amounts of upslope negative work once the crossfall increases from 0% (the blue ones) and those who continue to increase the upslope negative work applied as crossfall gradient increases.

As one would expect after seeing the high degree of variability in the data set, the data is not normally distributed, as shown by a Kolmogorov-Smirnov test ($p<.0001$). Therefore, a multiple linear regression was not run for this data and instead statistical significance of differences between the 3 groups (0%, 2.5% and 4%) was tested using a Friedman Test, with post-hoc analysis carried out using Wilcoxon Signed-Rank Tests . A Bonferroni correction[18] was applied, which resulted in a significance level of $p=.017$. This new significance level will be used to assess the effect of crossfall. However, the effect of occupant mass will not be tested.

There was a statistically significant difference in the amount of upslope negative work done by participants as crossfall increased ($X^2(2) = 21.571$, $p<.0001$). The Median (IQR) upslope negative work for 0%, 2.5% and 4% were -3.15,-13.38 and -12.61 respectively. There was no statistical significance between the upslope negative work done on the 2.5% and 4% crossfalls ($Z = -1.664$, $p=.096$). However, there was a statistically significant relationship

[18] A Bonferroni adjustment is used to correct for the increased chance of accepting a null hypotheses when it should in fact be rejected (a Type 1 error) when making multiple comparisons. To apply the Bonferroni adjustment the original alpha level (in this case p= .05) is divided by the number of planned comparisons. In this case 3 comparisons are being made (0 % against 2.5%, 2.5% against 4% and 0% against d 4%). (Pallant 2005)

between 0% and 2.5% ($Z = -3.296$, $p=<.0001$) as well as between 0% and 4% ($Z = -3.296$, $p=<.0001$). Therefore there is a significant effect of upslope negative work.

In summary, having looked at the provided capabilities needed to negotiate a footway with a crossfall it has been shown that a difference in work done on the upslope and downslope sides. The next section will address how people might apply the different forces necessary to keep a wheelchair moving, by examining in the push pattern used by the occupant.

5.6 Pushing Pattern

As defined in Section 3.3, the contact variables of interest to the research question are pushes, brakes and impacts. The number of pushes, brakes and impacts for each run were calculated and tested for normality using a Kolmogorov-Smirnov statistic, which showed none were normal ($p<.0001$). Therefore differences between the 3 groups (0%, 2.5% and 4%) were tested using three Friedman Tests (one for each contact type). Post-hoc analyses carried out using three Wilcoxon Signed-Rank Tests. A Bonferroni correction was applied to the post-hoc tests, which resulted in a significance level of $p=.017$.

The downslope results will now be presented (Section 5.6.1), followed by the upslope results. There is a slight difference in reporting structure between the downslope and upslope. In the downslope pushing patterns there is a far more detailed analysis of the number, magnitude and length of the pushes, the brakes are not analysed due to the fact there were virtually none and the impacts are only briefly discussed as although they existed in all conditions they did not change across crossfall gradients.

When reporting the upslope results (Section 5.6.2) a more balanced report of the three types of contacts is given as they each change with crossfall gradient.

5.6.1 Downslope Pushing Pattern

The number of pushes increased with crossfall gradient ($X^2(2) = 11.811$, $p=.003$), with the median number of pushes increasing from 8.5 on the 0% crossfall, to 10.2 and 11.7 on the 2.5% and 4% crossfalls respectively. However, when the differences between the three groups were tested it was found that the only significant difference existed between the number of pushes applied on the 0% condition compared with the number applied on the 4% crossfall ($Z = -2.671$, $p=.008$). This result can also be seen in the individual median values for each participant (see Table 5-12), which all show an increase from 0% to 4% (columns highlighted in green). However, there is no clear difference between 0% and 2.5% (Z = -1.570, p=.117) or between 2.5% and 4% (Z = -2.205, p=.027).

Table 5-12: Median number of pushes and push force.

	Median Number of Pushes			Median Push Force [N]		
Participant	0 %	2.5 %	4 %	0 %	2.5 %	4 %
1	12	15	17.5	24.5	31.3	42.3
2	6	8	12.5	64.9	54.2	59.1
3	8	11	14	49.2	56.0	60.9
4	5	8	10	68.6	66.7	79.0
5	7	7	10	68.8	75.3	75.4
6	8	13	10	47.0	35.5	49.1
7	9	9	10	43.6	43.6	57.0
8	10	10	15	32.6	30.8	56.3
9	5	5	6	68.3	77.9	102.8
10	12	12	12	47.6	51.6	70.5
11	12	11	17	31.4	31.4	48.6
12	11	11	14	40.0	40.0	39.4
13	6	6	8	79.1	85.0	96.2
14	6	6	8	53.5	53.5	67.9
Median (IQR)	8.5	10.2	11.7	56.7	54.0	60.6

Push Force was tested using a multiple regression model with crossfall and occupant mass as regressor terms (as the data was normal). The model was a poor fit, explaining just over 20% of variance in the data (R^2=.229, R_{adj}^2=.216), but was significant (F(2,122)=18.105, p<.0001). The constant term (15.03N) was found not to be significant (p=.065). However, both crossfall (β=3.7 N, p<.0001) and occupant mass (β=.508 kg, p<.0001) were significant (p<.0001).

When the median values for each of the participants are examined, see Table 5-12, it appears there is little difference between the 0% and 2.5% conditions. To see if this trend was statistically significant the push force needed on each of the three crossfall conditions was tested using a Wilcoxin Signed Ranks Test. The results of this test showed there was no significant difference between 0% and 2.5% (Z = -2.20, p=.826). However, there was a significant increase in force between 0% and 4% (Z = -3.170, p=.002) and between 0% and 2.5% (Z = -3.170, p=.005).

The number of brakes was not tested statistically on the downslope side as it was sporadic with a maximum value across all runs of 2, and with over 50 % of runs requiring no brakes at all.

The number of impacts did not increase significantly with crossfall gradient ($X^2(2) = 3.509$, $p=.173$), nor did the force applied during an impact ($X^2(2) = 1.000$, $p=.607$). The length of time the impact was applied for did increase significantly ($X^2(2) = 7.269$, $p=.026$), however, when the individual groups were tested for significance, there was no significant difference between any of the 3 groups.

In conclusion, the only type of contact to differ on the downslope as crossfall increased was the push. And although there was no difference between the 0% and 2.5% conditions, approximately 2 more pushes were required on the 4% crossfall compared with 0%. The magnitude of the peak push force also increased both with crossfall gradient and occupant mass.

5.6.2 Upslope Pushing Pattern

A summary of the median values for each contact type variable (average peak Ft, average contact time and average frequency) is given in Table 5-13.

Table 5-13: A summary of the median values for average peak Ft, contact time and frequency for each of the three contact types: Pushes, brakes and impacts, showing significant relationships when they exist according to the following key: ^ significant difference between 0% and 2.5%, * significant difference between 2.5% and 4%, $ significant difference between 0% and 4%, with significance level of p= .0017.

Independent Variable	0%	2.5%	4%
Push Ft [N]	46.18^	39.59^	38.27
Push Time [s]	.460	.425	.460
Push Freq [1/s]	1.21$^	1.05^	.995$
Brake Ft	-4.99	-13.18	-22.06
Brake Time [s]	.205	.400	.565
Brake Freq [1/s]	.04	.13	.23
Impact Ft [N]	-9.65$^	-16.00^	-15.26$
Impact Time [s]	.21$^	.54^	.59$
Impact Freq [1/s]	.85	1.18	.96

The push force used on the upslope side decreased significantly with crossfall ($X^2(2) = 7.000$, $p=.030$). However, post-hoc analysis revealed there was only a significant difference between 0% and 2.5% crossfalls ($Z = -2.856$, $p=.004$). Push frequency also decreased as crossfall increased ($X^2(2) = 7.000$, $p=.030$). Post-hoc analysis revealed push frequency decreased significantly from 0% to 2% ($Z = -2.480$, $p=.013$), as well as from 0% to 4% ($Z = -2.920$, $p=.004$). Push time did not significantly differ with crossfall gradient($X^2(2) = 1.057$, $p=.580$).

The amount of brake force decreased as crossfall gradient increased, however the trend was just below the significance level ($X^2(2) = 5.733$, $p=.057$). Average brake time and brake frequency were also not significant despite general increases in both time ($X^2(2) = 3.244$, $p=.197$) and frequency ($X^2(2) = 4.933$, $p=.085$).

The amount of force imparted to the handrim by the average impulse increased significantly with crossfall gradient ($X^2(2) = 18.143$, $p<.0001$). This increase was significant from 0% to both 2.5% ($Z = -3.296$, $p=.001$) and 4% ($Z = -2.731$, $p=.006$). The difference between 2.5% and 4% was not significant ($Z = -1.224$, $p=.221$). The average amount of time an impulse lasted also increased with crossfall ($X^2(2) = 15.164$ $p=.001$). Again this increase was significant from 0% to 2.5% ($Z = -3.041$, $p=.002$) and 4% ($Z = -2.982$, $p=.003$), but not between 2.5% and 4% ($Z = -1.259$, $p=.208$).The frequency with which impulses were performed did not change with crossfall gradient($X^2(2) = 2.714$ $p=.257$).

Overall it can be concluded that participants decreased the overall work done on the upslope side using two strategies. The first was to replace pushes with negative contacts. This can be seen by the decreasing frequency of pushes and the increasing frequency of negative contacts (both brakes and impulses). The second strategy was to make the negative contacts larger in both magnitude (Ft) and in length (contact time).

5.6.3 Conclusions

In conclusion when crossfall gradient increased the occupants changed their pushing pattern. They increased the number of pushes on the upslope side, and decreased the number of pushes on the upslope used; which were replaced with brakes. They also increased the duration of time spent on applying negative force to the handrim. Although there were general trends in the data, much of the data was not normally distributed, which was predominantly due to the variation between people. The high variation is due in part to the different strategies adopted by each person.

6 Attendant –Propulsion Results

This chapter reports the results of the experiments detailed in Chapter 0. In section 6.1 the details of the participants are reported, and in 6.2 the results of the maximum voluntary push force are given. The remainder of the chapter is divided into reporting the results of the required capabilities in the going phase (Section 6.4) are reported followed by the forces and velocities during the going phase(section 6.6), then the starting phase forces and stopping phase forces are reported (Section 6.6.1).

6.1 Participants

Fifteen participants took part in the study. Details of their personal characteristics are given in Table 6-1. There was a relatively even split between males and females, with 7 males and 8 females. The average age of participants was 66.53 (±3.76) years. The average weight was 73.18 (±8.39) kg, with females being on average only 0.9 kg heavier. The average height of participants was 171.6 (± 8.77) cm, with females on average 8 cm shorter. The majority of participants had no experience (5 people) or had pushed a wheelchair once or twice before (5 people); while 4 had had sporadic experience. No participant regularly pushed someone in a whelelchair.

Table 6-1: Participant details for self-propulsion experiments. Experience is measured on 4 point scale: 0= no experience, 1= have pushed a wheelchair once or twice, 2= sporadic experience and 3= regular (weekly) experience

Participant Number	Age [Years]	Weight [kg]	Height [cm]	Right/ Left Handed	Male (M) / Female (F)	Experience?
1	69	83.6	186	Right	M	1
2	65	63.35	167	Right	F	0
3	65	74.6	186	Left	M	2
4	65	58.3	166	Right	M	0
5	65	66.35	161	Left	F	2
6	69	76.10	173	Left	F	1
7	67	79.05	164	Right	F	2
8	68	79.5	174	Right	F	0
9	62	87.95	169	Left	M	2
10	62	81.0	166	Left	F	2
11	63	72.5	177	Right	M	0
12	75	75	178	Right	M	1
13	64	64	183	Right	M	1
14	66	66	164	Right	F	0
15	73	70.35	160	Right	F	1

6.2 Maximum Voluntary Push Force

The Maximum Voluntary Push Force (MVPF) test results are shown in Table 4-2. It can be seen there was an average MVPF of 223 N (± 70 N). The amount of resultant force (F_{res}) was only slightly higher than that of the MVPF, with a mean of 235 N (74 N).

Table 6-2: Results of MVPF test for attendants

P#	Max F_{hor}	Max F_{res}
1	235.87	241.56
2	156.48	158.18
3	405.26	434.11
4	149.79	163.24
5	183.30	197.79
6	220.99	242.37
7	218.29	233.85
8	225.93	252.08
9	208.90	212.45
10	244.16	256.46
11	337.37	343.78
12	184.88	208.24
13	128.13	132.48
14	220.30	223.17
15	228.65	231.07
Average	223.22	235.39
Std.Dev	70.20	73.99

6.3 Deviation from a straight line

No participants deviated noticeably from a straight line during these experiments. This shows that the provided capabilities of the APWS were greater than the required capabilities imposed by the environment.

6.4 Provided capabilities of going Phase

The provided capabilities can be expressed in terms of the amount of work required to propel the wheelchair along a crossfall. As the interface between the attendant and the wheelchair is in the form of two handles, the sum of the work from the left and right handles represents the overall work done on the wheelchair system. If no work is done overall, then the wheelchair will simply stay put. The task was to push a wheelchair in a straight line. We know that all participants were able to achieve this (see Section 6.3), the next step is to discover which provided capabilities they used in achieving this. This will be done by investigating the sum of the work (C_{wk_sum}) and Difference of Work (C_{wk_diff}) will now be examined for the going phase.

6.4.1 Provided going Work & Provided going Work Difference

The results of the regression model for C_{wk_diff} and C_{wk_sum} using crossfall gradient as the regressor term are given in Table 6-3.

Table 6-3: Regression Model Summary for the provided going work (Cwk_sum) and the difference of work (Cwk_diff).

Dependant Variables	Model		Coefficients (results of t-test)			
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P
C_{wk_diff}	.745 (.742)	<.0001	2.43	.718	36.77	<.0001
C_{wk_sum}	.000 (-.013)	.952	194.50	<.0001	-.126	.952

Under the column headed 'Model' are the R^2 and R_{adj}^2 values for each of the variables. For C_{wk_diff} the model has a good degree of fit; accounting for approximately 74% of the variance found in C_{wk_diff}. The model is also statistically significant ($F(1,79)=13.866$, $p=<.0001$). The coefficients of the model were assessed with two separate t-tests, the results of which are shown in the four columns to the right in Table 6-3. It can be seen that the constant term is not significant, which is probably due to the large degree of variation between people. This variation is also visible in Figure 6-1, which shows the individual run values for C_{wk_diff} and the results of the regression line using the coefficients from Table 6 3. The coefficient for crossfall, which constitutes the slope of the regression line in Figure 6-1 was statistically significant ($p<.001$); an additional 36.77 N was provided for each percentage increase in crossfall gradient.

The results of the regression analysis for C_{wk_sum} with crossfall as the regressor term are also

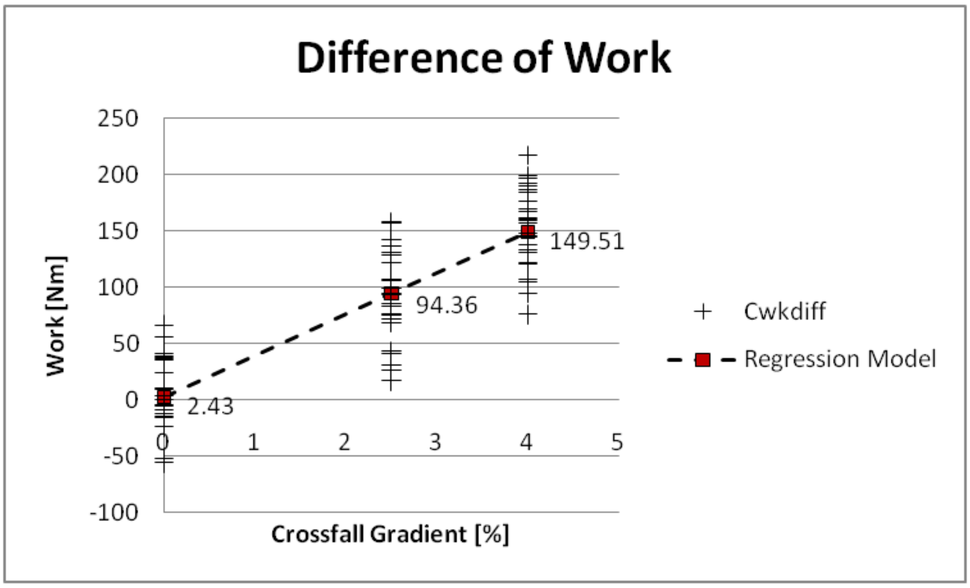


Figure 6-1: Provided Going Work Difference

shown in Table 6-3 (row 2). Under the column entitled 'Model' we can see that the R^2 value is positive, while the R_{adj}^2 is negative. This indicates the model is not able to predict a linear trend in the data that significantly differs from zero. This is backed up by the significance level of the model, which is far larger than .05 ($F(1,79)=.004$ $p=.952$). It is further qualified by the fact that only the coefficient for the constant term is statistically significant ($p<.0001$) and there is no effect of crossfall ($p=.952$).

Figure 6-2 shows the individual values for each run for C_{wk_sum}, along with the regression line found using the coefficients from Table 6-3, which is virtually a straight line.

It can therefore be concluded that the amount of C_{wk_sum} needed to push along a footway is

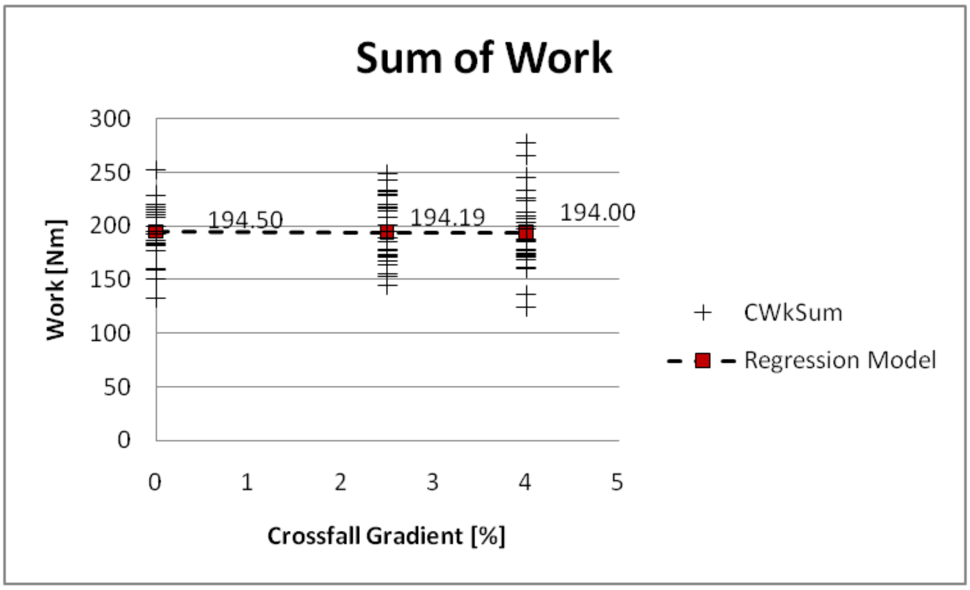


Figure 6-2: Sum of Work for the APWS

independent of crossfall gradient. Therefore, the null hypothesis for C_{wk_sum} cannot be refuted. However, the amount of difference of work required is statistically significant, and therefore the null hypothesis can be rejected. For this reason we can say that there is an increase in the capabilities required to push a wheelchair on a footway with a crossfall gradient of greater than 0%, compared to one of 0%. In fact it increases approximately 36 Nm with each percentage increase in crossfall. How people produce this difference of work necessary to continue along a straight line with a crossfall will now be investigated by examining the downslope and upslope work.

6.4.2 Downslope Positive Work

Downslope work is defined as the work done on the handle which is on the downslope side, while upslope work is defined as the work done on the handle which is on the upslope side. For the 0% crossfall, 'downslope' was defined as the dominant hand of the participants and 'upslope' the non-dominant hand.

It can be seen in Figure 6-3 that there is a general trend of increased downslope work; and in Figure 6-4 that there is a general decrease in upslope work. However the variation between people and runs is high, as can be seen in the spread of data for each crossfall condition in Figure 6-3 and Figure 6-4. This is particularly the case for upslope work (Figure 6-4).

Despite the large spread in the data, the results of the linear regression models for upslope work ($F(1,79)=150.51$, $p<.001$) and downslope work ($F(1,79)=116.22$, $p=<.001$) are significant. The model for upslope work is capable of accounting for 65 % of the variance in upslope work, and the model for downslope work is able to explain 59 % of the variance recorded in that variable. The results of the models for both upslope work and downslope work are given in Table 6-4.

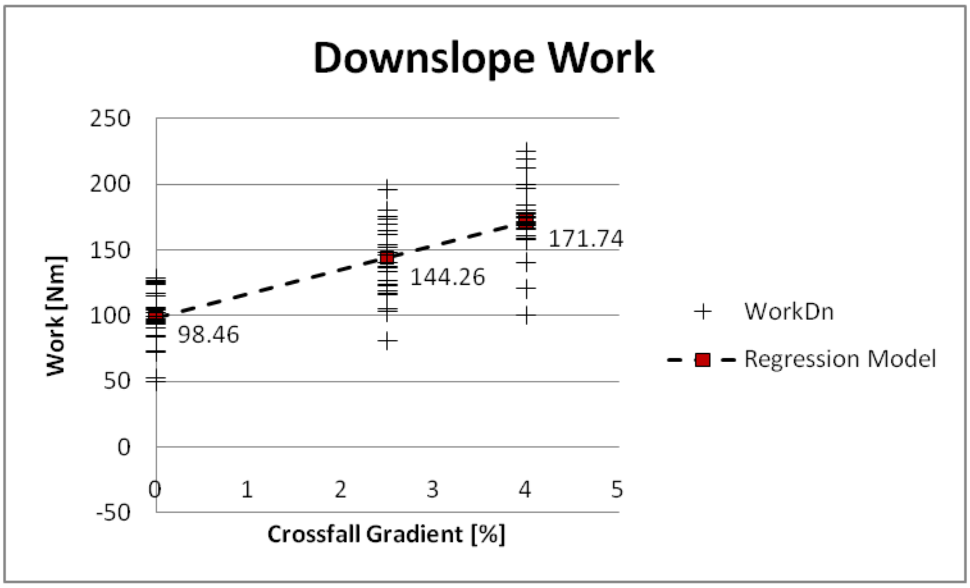


Figure 6-3: Downslope Work for APWS

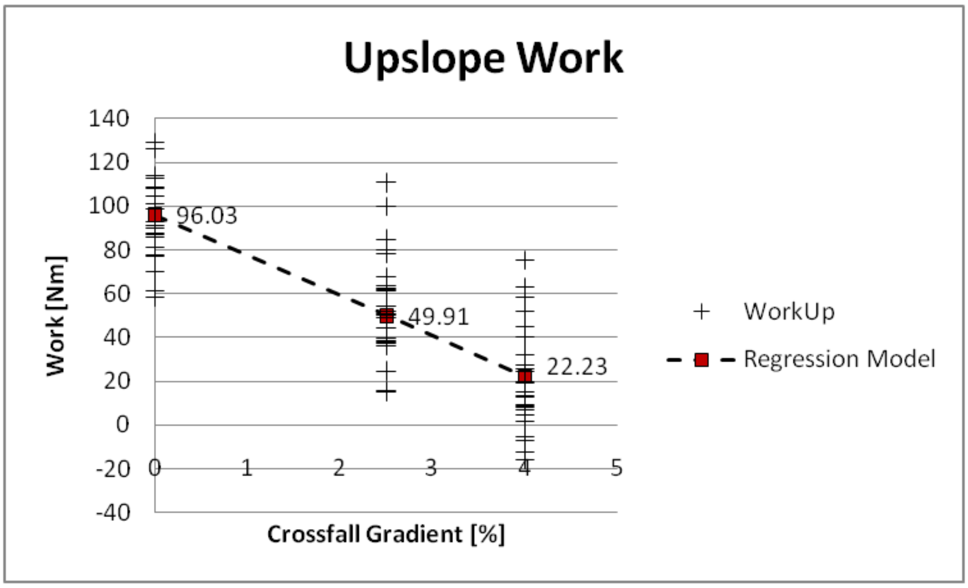


Figure 6-4: Upslope Work for APWS

In Table 6-4 the constant terms for both downslope work (row 2) and upslope work (row 1) are very similar, 98.46 Nm and 96.03 Nm respectively. Both terms are also significant ($p<.0001$). However, downslope work is shown to increase with crossfall gradient (18.32 Nm per percentage increase), while the upslope work decreases (-18.45 Nm per percentage increase). Both coefficients are again significant ($p<.001$). When both downslope work and upslope work are taken together there is a difference of work per percentage increase in crossfall of 36.77 Nm, which accounts for the difference reported to be needed in section 4.4.1 (page 101).

Table 6-4: Regression Model Summary for the downslope and upslope work

Dependant Variables	Model		Coefficients (results of t-test)			
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P
downslope work [Nm]	.595 (.590)	<.001	98.46	<.001	18.32	<.001
upslope work [Nm]	.656 (.651)	<.001	96.03	<.001	-18.45	<.001

6.4.3 Positive and Negative Work

Downslope work is calculated by summing the negative downslope work (caused by a push force of less than zero on the downslope handle) and positive downslope work (caused by a Push force greater than zero on the downslope handle). Likewise, the upslope work is

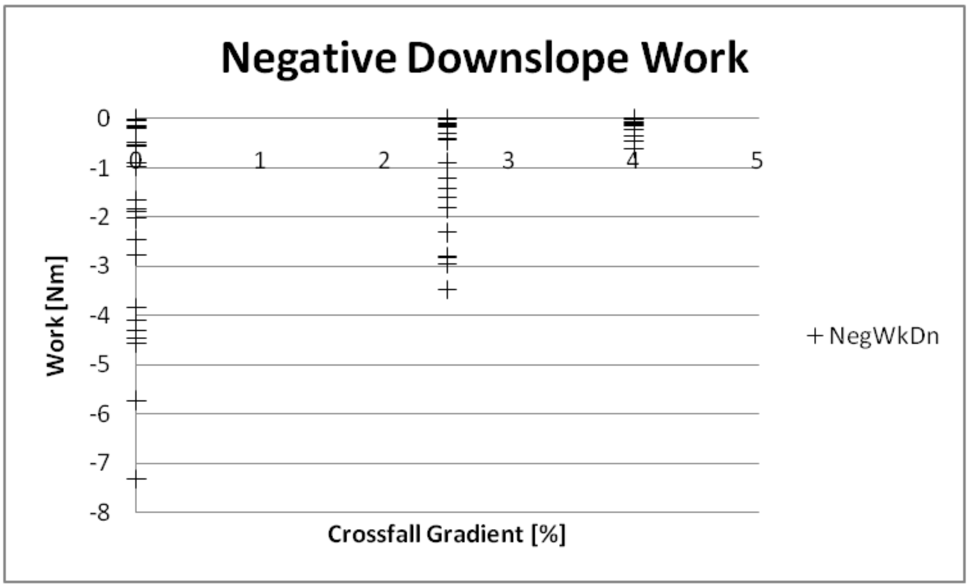


Figure 6-5: negative downslope work for APWS

calculated by summing the negative upslope work and the positive upslope work. These four parameters will now be investigated, starting with the downslope negative work.

6.4.4 Downslope Negative Work

There is very little negative downslope work for any run. This can be seen in Figure 6-5, where each individual run has been plotted. The negative downslope work decreases with crossfall gradient from a very small amount on the 0% to the point where there is virtually none on the 4% crossfall.

Due to the very small numbers involved in the Downslope Negative Work, the downslope positive work (see Figure 6-6) is nearly identical to the downslope work (see Figure 6-3). For completeness the results of the linear regression model for downslope positive work are given in Table 6-5. The coefficients are very similar to those found for downslope work, as would be expected given the limited values of downslope negative work, and the model was significant ($F(1,79)=100.53$, $p<.001$).

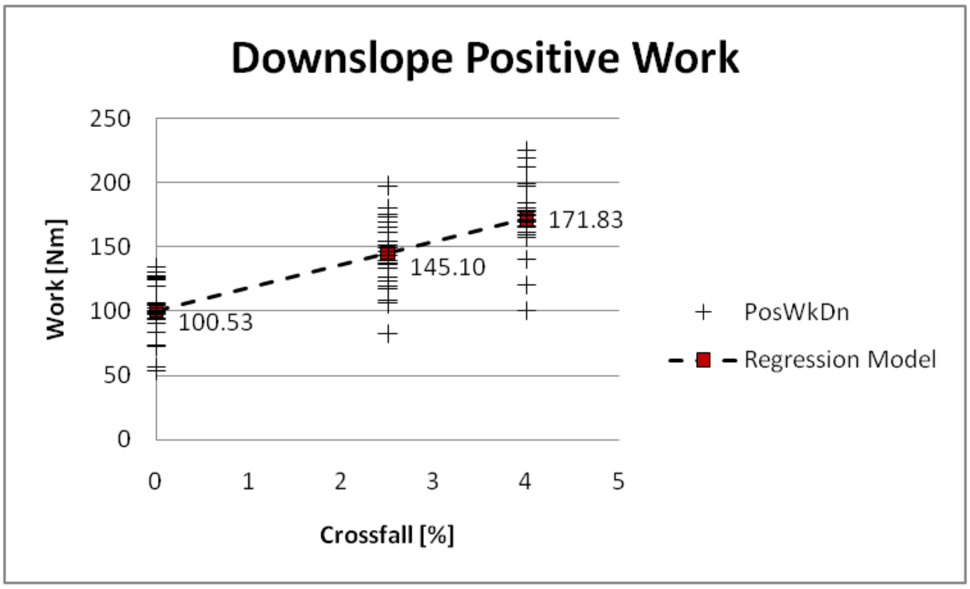


Figure 6-6: Downslope Positive Work for APWS

It can be concluded that virtually all the increase in downslope work is the result of the increase in downslope positive work. The results of how the upslope work was divided between upslope positive work and upslope negative work will now be investigated.

Table 6-5: Regression Model Summary for the Downslope Positive Work

Dependant Variables	Model		Coefficients (results of t-test)			
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P
Downslope Positive Work [Nm]	.586 (.581)	<.001	100.53	<.001	17.826	<.001

6.4.4.1 Upslope Positive and Negative Work

In section 6.4.2 it was found that there was a reduction in upslope work done as crossfall gradient increased. This section investigates if this reduction was the result of the attendants pulling on the upslope handle and thus creating upslope negative work, or just pushing less hard, which would result in a reduction in upslope positive work. The results of the linear regression model for upslope positive work and upslope negative work with crossfall gradient as the regressor term are shown in Table 6-6.

Table 6-6: Regression Model Summary for the Upslope Positive Work and Upslope Negative Work.

Dependant Variables	Model		Coefficients (results of t-test)			
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P
Upslope Positive Work [Nm]	.621 (.616)	<.001	97.53	<.001	-14.15	<.001
Upslope Negative Work [Nm]	.485 (.478)	<.001	-1.50	.281	-4.30	<.001

The model for the amount of upslope positive work was a good fit accounting for 62% of the variance (F(1,79)=129.57, p<.001). The effect of crossfall was found to be significant (p<.001) when tested with a t-test; with a negative coefficient of -14.15 Nm. Thus, the amount of upslope positive work accounts for nearly the full -18.45 Nm decrease found in upslope work (section 6.4.2, page104).

The remainder of the upslope work decrease is as a result of a decrease in upslope negative work. This is shown in the results of the linear regression model for upslope negative work (Table 6-6, row 2). The model had a reasonable fit and was able to explain 48% of the variability (F(1,79)=74.35, p<.001). The constant term was not significant when tested with a t-test (p=.281). However, there was a significant decrease in upslope negative work as represented by the negative coefficient for crossfall in Table 6-6. For each percentage increase in crossfall gradient there was a -4.3 Nm decrease in upslope negative work.

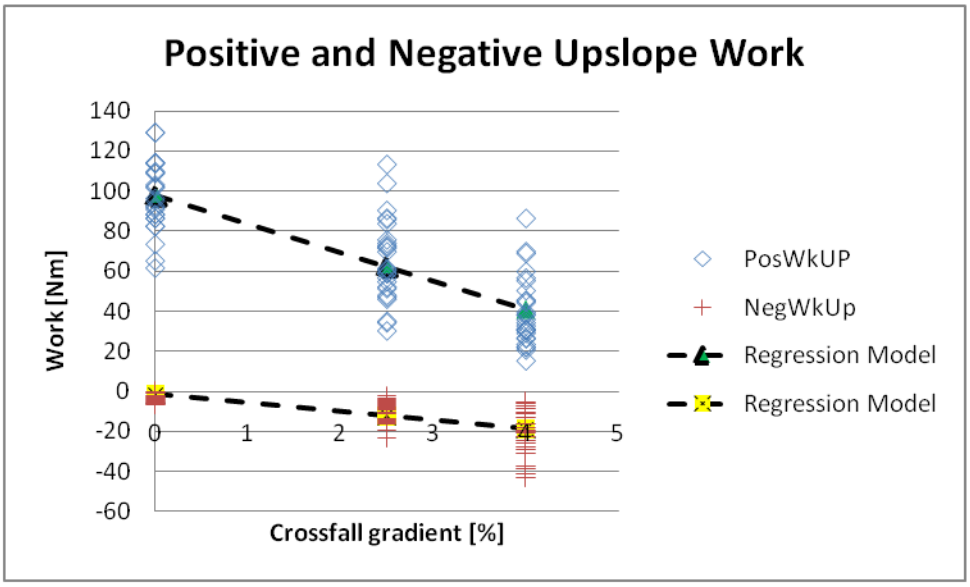


Figure 6-7: Positive and Negative Upslope Work showing regression lines using coefficients in table 5.

It can be concluded that both the upslope positive work and the upslope negative work decrease significantly with crossfall gradient. Therefore the null hypothesis in each case can be rejected.

Despite this overall trend, when the individual values of upslope negative work and upslope positive work are plotted for each participant (Figure 6-8), there is a clear difference in the amount of upslope positive work done by participant 13 when compared to the other 6 participants.

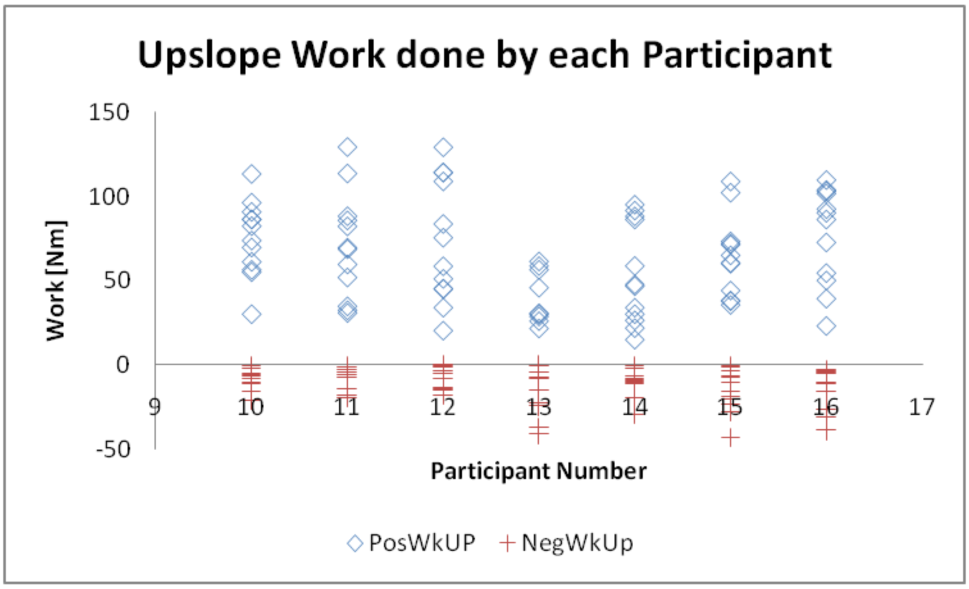


Figure 6-8: Upslope work done by each participant, showing rather low values of Upslope Positive Work for participant 13.

Participant 13 also happened to be the oldest of the participants (see Table 6-1). One way he may have been able to do this would have been to push the wheelchair more slowly, which would then have required less Push Force (the horizontal component of the force applied to the handles). These 2 parameters (Velocity and Push Force) will now be investigated.

6.5 Forces and Velocity

Figure 6-9 shows the average velocity of the individual runs for each of the 7 participants discussed thus far. It is clear that participant 13 has chosen to travel more slowly than the other six participants, which accounts for the reduced upslope work values found in section 6.4.4.1.

6.6 Push Forces

Attendants are able to create positive work by applying a Positive Push Force to the handles, while they can create a negative work by applying a Negative Push Force to the handles and in effect pulling on the handle. It would be expected that the results showing an increase in

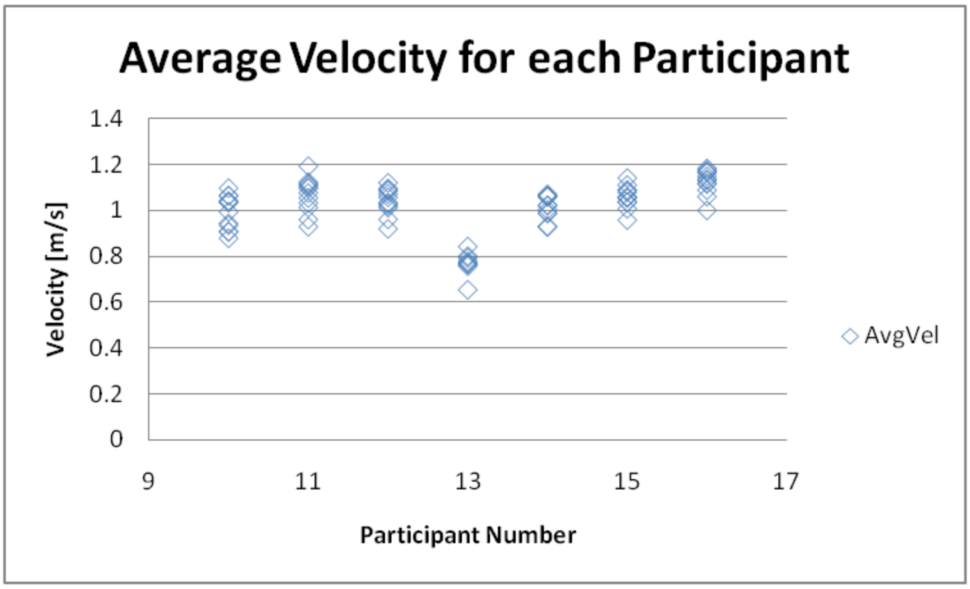


Figure 6-9: Average of left and right rear wheels velocities during the going phase, showing participant 13 chose to travel more slowly than the other participants.

downslope positive work reported in section 6.4.4 would be the results of an increase in Downslope Push force, and the decrease in Upslope Positive Work and of upslope negative work would be the result of a decrease in Upslope Positive Push Force and Upslope Negative Push Force respectively. These Push force parameters are now reported.

When the Upslope Push Force and Downslope Push Force are plotted against crossfall gradient (Figure 6-10), a general trend appears of increasing Upslope Push Force and decreasing Downslope Push Force. These trends are confirmed by the linear regression models for Upslope Push Force and Downslope Push Force.

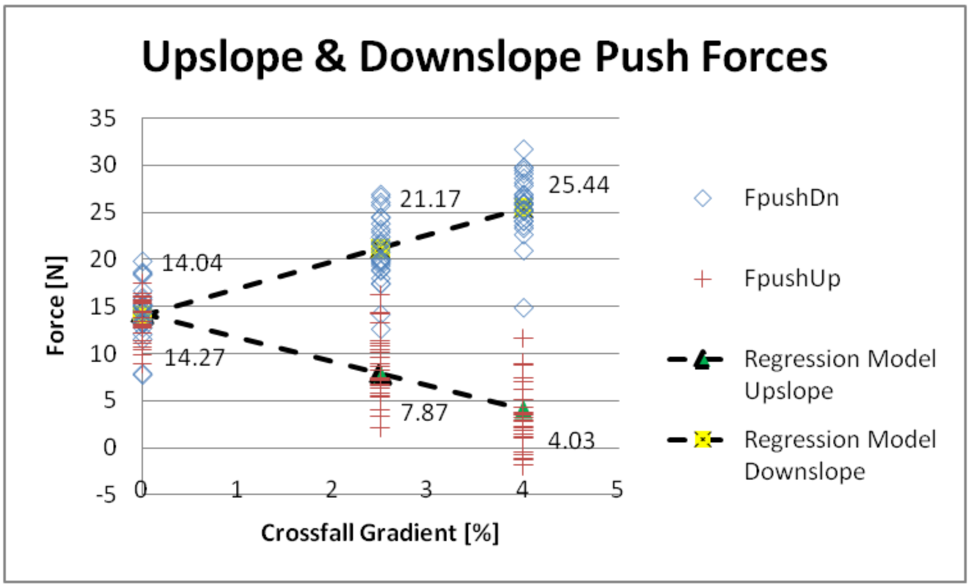


Figure 6-10: Upslope & Downslope Push Forces

The results of the linear regression models are shown in Table 6-7. The trend of increasing Downslope Push Force was significant (F(1,79)=167.84, p=<.001) as was the tendency of decreasing Upslope Push Force (F(1,79)=141.68, p=<.001). The constant terms were very similar for both variables; 14.04 N for the Upslope Push Force and 14.27 N for the Downslope Push Force. The magnitudes of the coefficient for crossfall gradient for the Upslope Push Force (-2.56 N) and the Downslope Push Force (2.85 N) were also very similar, though they differed in sign (see Table 6-7).

Therefore it would appear people adapted to the crossfall by applying equal magnitudes of increased Downslope Push Force and decreased Upslope Push Force.

Table 6-7: Regression Model Summary for the downslope and upslope work.

Dependant Variables	Model		Coefficients (results of t-test)			
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P
Upslope Push Force	.642 (.637)	<.001	14.04	<.001	-2.56	<.001
Downslope Push Force	.680 (.676)	<.001	14.27	<.001	2.85	<.001

6.6.1 Starting & stopping

The peak starting and stopping forces on the upslope and downslope sides of the wheelchair will now be reported.

The mean values of the peak starting and stopping forces are show in Table 6-8 and the results of the regression analysis shown in Table 6-9. The Downslope Starting Force and Upslope Starting Force are nearly identical for the 0 % crossfall and then begin to diverge as the crossfall increases (see Table 6-8).

Table 6-8: Mean values of Starting and stopping peak forces for each crossfall

Independent variable	0%	2.5%	4%
Downslope Starting Force [N]	50.35	55.45	58.80
Upslope Starting Force [N]	50.30	49.84	45.34
Downslope stopping Force [N]	-36.44	-28.32	-26.35
Upslope stopping Force [N]	-38.15	-39.89	-48.65

There was a significant increase in Downslope Starting Force ($p<.001$). However, the model was a poor fit, explaining approximately 14% of the variation in the data and in fact the effect of crossfall gradient only caused an increase of around 2 N per percentage increase in crossfall gradient (see Table 6-8, row 1). While it is significant, and the null hypothesis can be refuted, the actual force increase is less than 5 % of the push force provided to start the wheelchair in motion.

The Upslope Starting Force reduced by a small amount as crossfall increased. However, the decrease was not significant and therefore the null hypothesis cannot be rejected (see Table 6-8, row 2).

Table 6-9:Regression model parameters for the starting and stopping forces for the upslope and downslope handles of the wheelchair

Dependant Variables	Model		Coefficients (results of t-test)			
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P
Downslope Starting Force [N]	.146 (.135)	<.001	50.30	<.001	2.11	<.001
Upslope Starting Force [N]	.044 (.031)	.061	51.02	<.001	-1.15	.061
Downslope stopping Force [N]	.131 (.120)	.001	-35.95	<.001	2.58	.001
Upslope stopping Force [N]	.044 (.031)	.009	-36.84	<.001	-2.47	.009

The Downslope stopping Force significantly increased (less negative force was applied) with crossfall gradient ($p<.001$). However, the model was a poor fit accounting for around 12 % of the variance found in the data.

The Upslope stopping Force significantly decreased at approximately the same rate as the Downslope stopping Force increased ($p<.009$). However, the model was an extremely poor fit, representing only 4% of the variance, and so in effect, not useful.

In summary, the peak forces used to start the wheelchair were higher than those used to stop the wheelchair and on the 0% condition the forces applied to either side were very similar. When a crossfall was introduced, people provided approximately equal and opposite forces to the upslope and downslope handles for both stopping and starting. This means there is a moment being created be the attendant, which is overcoming the crossfall, this creates an asymmetric push force for the attendant. Therefore there is a need for attendants to be able to provide a difference of force when starting and stopping, and the magnitude is approximately the same as that provided during the going phase; approximately 6 N (see Table 6-7 and Table 6-8).

7 Discussion

Chapter 5 reported the results of the provided capabilities and Coping Strategies used by a Wheelchair occupant when traversing a crossfall. Chapter 6 reported the results of the provided capabilities of attendants when pushing a wheelchair along a crossfall. This chapter first states the difference in provided capabilities needed by both types of users when traversing a crossfall greater than 0 % compared with 0% (Section 7.1), it then discusses the various types of Coping Strategies used by attendants (Section 7.3) and compares the forces used by attendants to manual handling guidelines set for pushing and pulling (Section 7.2.1). The results of the self-propulsion wheelchair experiments are then discussed focusing firstly on the coping strategies of occupants (Section 7.3.1), then comparing the cases of participant 1 and participant 14 (Section 7.3.2). Then the definition of the pushing pattern is revisited and a visual aid to describing the effort needed to push a wheelchair proposed (Section 7.3.3). Finally crossfalls are compared to other barriers faced by wheelchair users (Section 7.3.4).

7.1 Provided Capabilities Needed to Traverse Crossfalls

The provided capabilities for the APWS and the SPWS increase when the crossfall gradient increases from 0 %. When the footway is flat there is a single provided capability for each of the three phases (starting, going and stopping) of manoeuvring a wheelchair. However, when the crossfall gradient increases to 2.5% or 4% a second provided capability is needed in order to keep the wheelchair travelling in a straight line. This second provided capability necessitates the user to apply a difference of force to the upslope side of the wheelchair compared to the downslope side. This must be done while conserving the total amount of force applied to the wheelchair. How this was achieved differed depending on the person and the type of propulsion. The discussion will now turn to how attendants achieved this provided capability, where the forces necessary to accomplish it will be compared to manual handling guidelines. Following this the strategies employed when self-propelling will be classified and crossfalls ranked in terms of other barriers in the built environment

7.2 Attendant-Propulsion

As shown in Chapter 6, none of the attendants tested in these experiments had a problem maintaining a straight line while pushing the wheelchair and all claimed it to be ‘fine’ while completing the experiments. However, many did appear to become out of breath on occasion.

During attendant-propulsion the attendant is constantly applying some amount of force to the wheelchair, and while this is of benefit when combating a barrier such as a crossfall, it also does not allow any period of rest for the attendant. The way force is imparted to a wheelchair by an occupant, in contrast, does allow for periods of rest as the wheelchair coasts between pushes.

Attendants reduced the amount of work done on the upslope side (Figure 6-4) while simultaneously increasing the amount of work done on the downslope side (Figure 6-3). However, this decrease in upslope work contained only very small amounts of negative force, and in fact there was a greater amount of negative force applied on the flat condition than on the positive crossfalls. It is suggested here that due to the natural sway from left to right (medial-laterally) during a gait cycle, there is an inherent amount of push force and then pull force during a gait cycle, and the fact that this is reduced with increasing crossfall would suggest a change in gait pattern or a stiffening of the upper body, which prevents this sway revealing itself in the force curves. Unfortunately, gait cycle characteristics were not investigated in this thesis.

As has just been stated, as crossfall gradient was increased, attendants did not apply an increased negative force to the upslope handle. Therefore it remains unknown if the participants had the Provided Capability (of being able to apply a positive force to one handle while applying a negative force to the other, whilst walking) and chose not to use it, or simply do not possess this Provided Capability. It may be the case that this strategy is only employed on steeper crossfalls than those tested in this thesis, or when the amount of push force being applied on the downslope side begins to approach their MVPF strength.

One participant in the attendant-propulsion experiments (P13) went noticeably more slowly than all the others by approximately 0.3 m/s, travelling with an average velocity of 0.8m/s. This was despite having a MVPF result of 218 N, which was slightly higher than the average of 209 N. As the sample size of this experiment is particularly small it is difficult to infer anything from this for example, it could simply be a result of personal preference. However, it could also be indicative of an increased difficulty in applying push force as people age (this participant was 75 years old at the time of the experiment), which has been reported previously (Voorbij & Steenbekkers 2001). However, it is difficult to establish this for certain without longitudinal testing of individuals on such tasks over time. The issue may not be about the effect of age per se as much as the effect of age on an individual, and to have a true picture it would be necessary to compare a person's performance over time.

7.2.1 Crossfalls Relative to Manual Handling Guidelines

It is assumed in these guidelines (see Pg. 30) that approximately equal forces are being applied by the left and right sides. Therefore, when comparing to the upslope and downslope values reported in this thesis they have be halved.

7.2.1.1 Starting Force

The mean value for the Downslope Starting Force on the 4% crossfall was 58.8 N, and the maximum value recorded was 77.9 N. In general the Starting Forces were well within the manual handling guidelines for all the male participants when the crossfall was 0% (see Figure 7-1). There was a single trial where a female participant exceeded the guidelines for the 0%

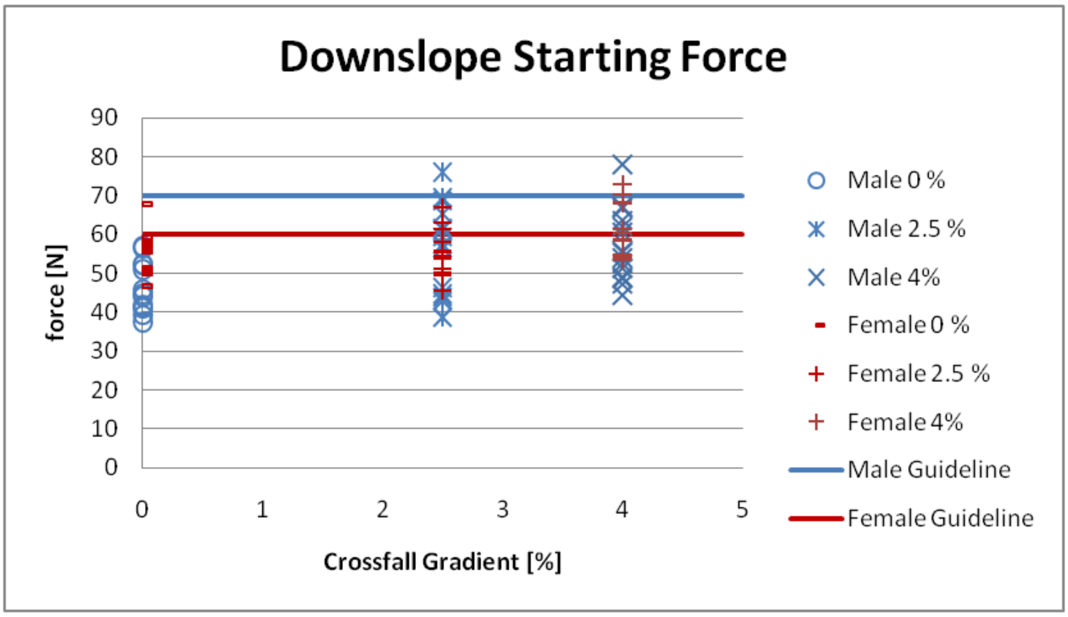


Figure 7-1: Downslope Starting Forces plotted against crossfall gradient, showing the guidelines for peak initial forces when pushing 45 m for males and females as recommended by Snook & Ciriello 1991.

crossfall (see Figure 7-1). As crossfall gradient increased so did the frequency with which the guidelines were exceeded both in the cases of males and females (see Figure 7-1).

7.2.1.2 Stopping

The guidelines for the initial peak for pulling recommended by Snook and Ciriello (Snook & Ciriello 1991) were taken for comparative purposes with the stopping Forces. The upslope starting and stopping forces were higher than those on the downslope, and so these have been assessed against the guidelines. As was the case for the Starting Forces the Stopping Forces on the upslope side are well within the guidelines set by Snook and Ciriello for males and also in this case for females. However, they creep closer to the maximum guideline as crossfall increases and in two instances exceed the guideline (see Figure 7-2).

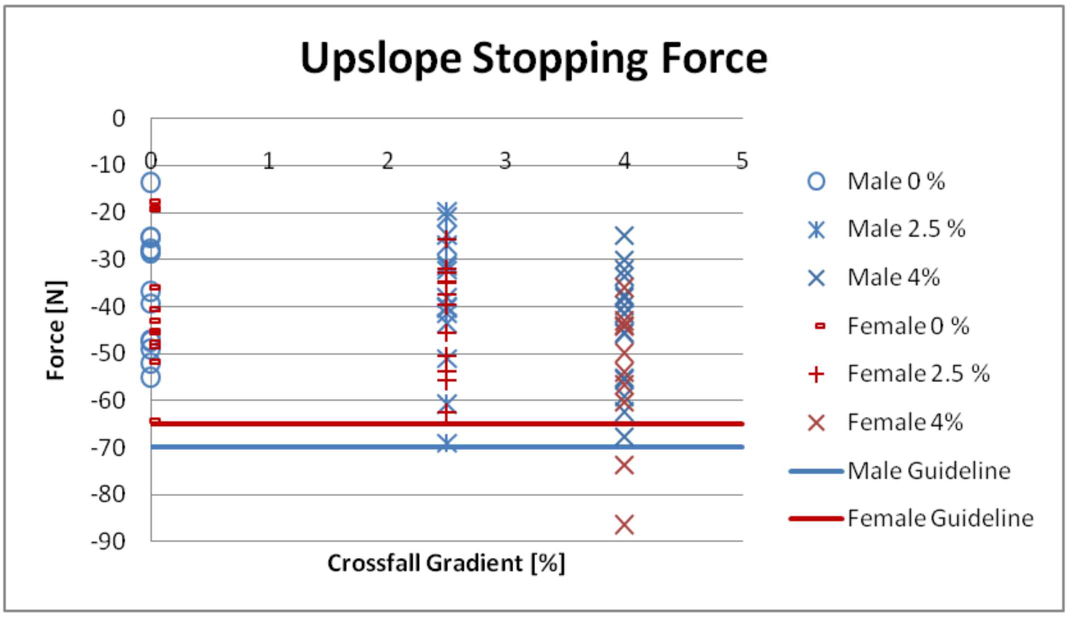


Figure 7-2: Upslope Stopping Forces plotted against crossfall gradient, showing the guidelines for peak initial forces when pushing 45 m for males and females as recommended by Snook & Ciriello 1991.

7.2.1.3 Going

There was shown to be a trend of increasing force on the downslope side relative to the upslope during the going phase. When pushing on the 0% crossfall the forces used by attendants are far lower than the guidelines state as a maximum (see Figure 7-3). However a few males and the nearly half the females exceed the maximum guideline when pushing on a crossfall gradient of 2.5 % and 4 % respectively (see Figure 7-3).

In summary, the introduction of the additional Provided Capability when pushing on a crossfall necessitates that attendants exceed guidelines set for industrial manual handling limits. It should be noted that these guidelines were constructed based on data from health young individuals and many carers may not fit this profile. It might, for example, be the case that 'guidelines' for older people, if they were to be produced, would indicate lower maximum forces.

In the same vein if 'guidelines' were produced for attendant wheelchair propulsion they might result in lower force values due to the increase in distance (from the 45m measured by Snook and Ciriello) travelled when pushing someone for the day in a wheelchair.

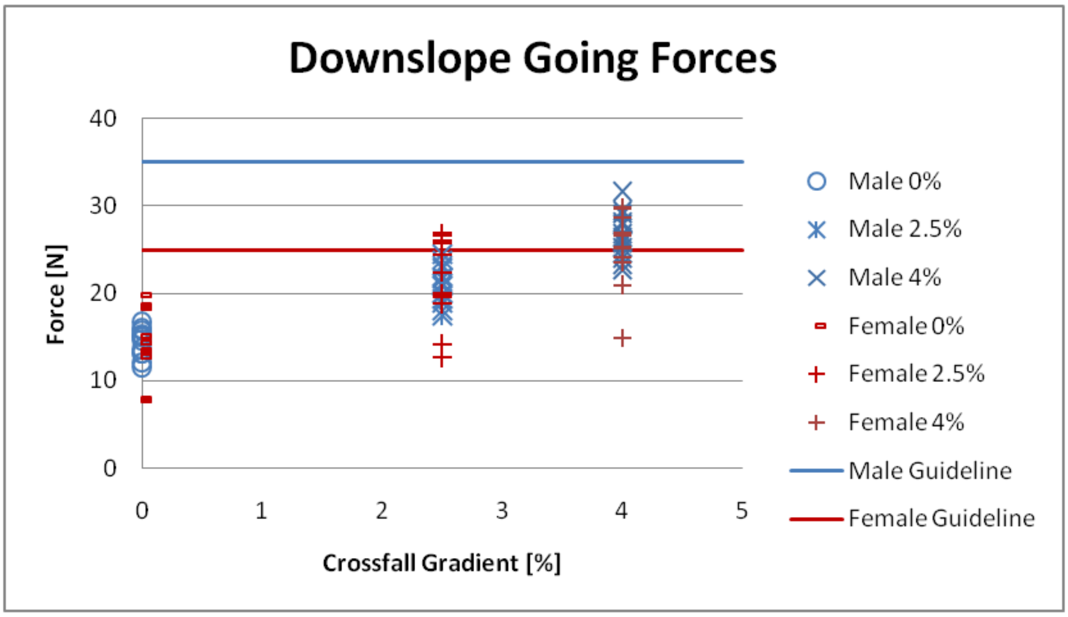


Figure 7-3: Downslope going forces plotted against crossfall gradient, showing the guidelines for peak initial forces when pushing 45 m for males and females as recommended by Snook & Ciriello 1991.

7.3 Self-propulsion

7.3.1 Coping Strategies

There were a number of different strategies employed by occupants to overcome the required capabilities of the crossfalls. These are shown in the results in Section 5.6 and will be examined in more detail here. It appears there were 4 different strategies employed by occupants to move the SPWS.

The first strategy was to reduce the amount of force applied to the upslope handrim. This is illustrated in Figure 7-4 (top row), in which the reduction in upslope forces can be seen as the crossfall increases.

The second strategy is to apply more force to the downslope handrim (see Figure 1 bottom row, in which the increase in forces can be seen as the gradient increases). Each of these strategies could allow the occupant to maintain a regular pushing pattern, while maintaining their desired velocity (see Figure 7-4). However, they could be used in combination. It was not possible on this occasion to obtain simultaneous recordings of downslope and upslope forces so the impact of such a combined strategy could not be analysed in this study. Although the two forces were analysed separately, the task was completed in a way so that the participant was in ignorance of which wheel was actually recording the forces. This means that it is safe to assume that the forces being recorded in each case are actually the forces resulting from the coping strategy and take into account the combination of force inputs rather than just the single input on one side of the wheelchair. This was also confirmed by watching the videos of the experiments in detail.

It is assumed given previous research that the first strategy would reduce upper limb loading, whereas the second strategy would increase this loading (Boninger et al 2004b; DiGiovine *et al.* 1997). Simultaneous use of both strategies might reduce the load in each case compared with the application of only one of these strategies, but could add the need to be able to manage the simultaneous differential application of forces. For completeness, it would be useful to undertake simultaneous recording of both sides in a subsequent experiment in order to clarify this point.

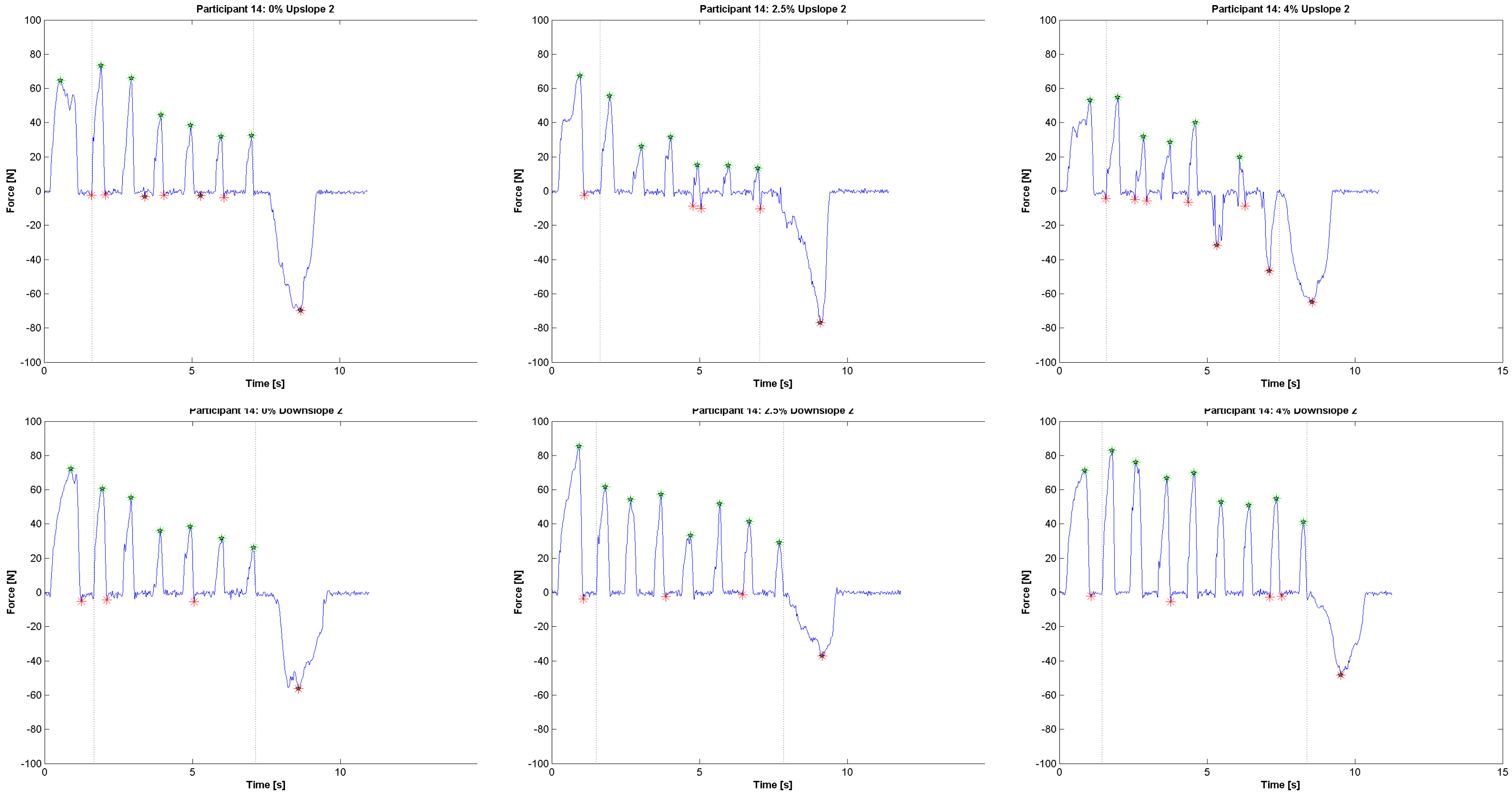


Figure 7-4: Tangential force data from Participant 14. The top row shows the upslope runs and the bottom the downslope runs. There is a decrease in force as crossfall gradient increases (read from left to right) on the upslope side, whereas there is an increase in force on the downslope side. The vertical dashed lines represent the start and end of the Going phase. The peak forces of each contact are highlighted with the following key: the black and red stars are Brakes, green stars are Pushes and red stars are Impacts.

An extreme case of reducing the upslope force is to apply virtually none as was the case when P13 traversed the 2.5% crossfall (see Figure 7-5). For all 3 runs P13 imparted only one Push on the handrim, which was during the Starting Phase and then applied no further Pushes. However, P13 did apply a couple of small Brakes during the going phase and then a large Brake during the stopping phase. Interestingly P13 did not apply greater force on the downslope side to counter their chosen strategy on the upslope side. Instead they chose to travel more slowly, which agrees with the general trend seen in Richter *et al.*'s study of reduced velocity with increasing crossfall. (Richter *et al.* 2007a).

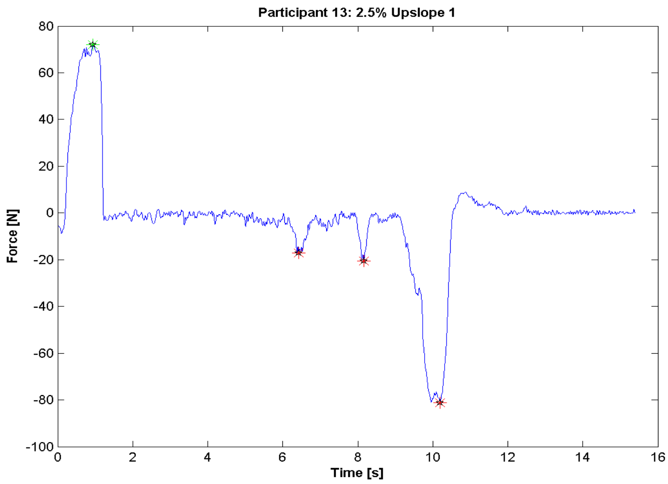


Figure 7-5: Extreme example of the strategy to reduce Going Work on the upslope side.

The third strategy is to apply negative force to the upslope side of the wheelchair; this can be done by increasing the number, magnitude or duration of Impacts on the handrim, or increasing the number, magnitude or length of the brakes compared with the downslope side (Figure 7-6).

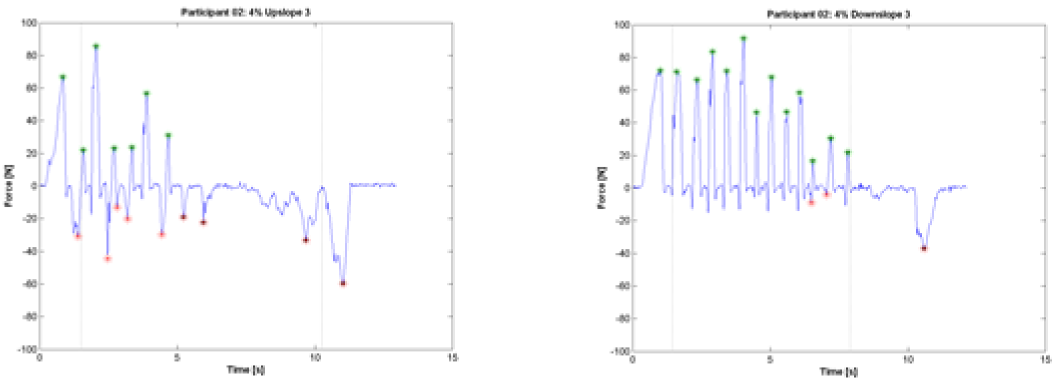


Figure 7-6: Example plots from Participant 2 showing an increasing number of impacts (red stars) on the upslope side (left) compared with the downslope side (right)

Neither of these strategies is mechanically efficient as Brakes and Impacts do not directly translate into motion in the desired direction, even if they do assist the occupant in controlling the direction of travel. Despite not directly contributing to forward motion, these contact types do require the occupant to apply a force, and therefore they are doing work, even if there is no direct increase in the work done by the SPWS.

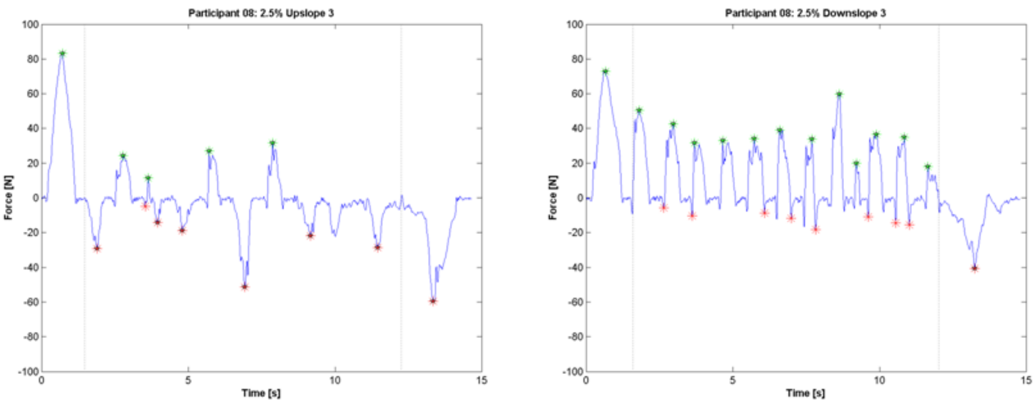
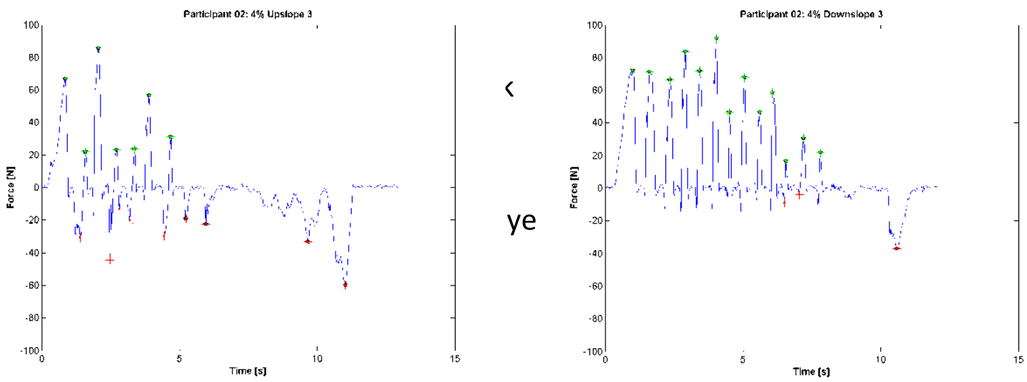


Figure 7-7: Example plots from Participant 8 showing an increasing number of brakes (red stars) on the upslope side (left) compared with the downslope side (right)

In summary, there are 4 coping strategies to overcome a crossfall when self-propelling:

1. Reduce upslope work by decreasing positive work
2. Increase downsl
3. Reduce upslope
4. Travel more slow



These can be used indep ye

7.3.2 Comparison of P1 and P14

In order to understand more about the processes involved in self-propulsion on a crossfall, it is instructive to compare two participants who are broadly similar in terms of age, height, weight and gender but who produce very different performance characteristics. In broad terms, Participant 14 seems to have a well-controlled style of pushing the wheelchair, whereas Participant 1 is much more erratic. Comparison of these participants therefore could shed some light on the effects (or otherwise) of irregular as opposed to regular propulsive performance.

Participant 1 struggled to complete the task when the crossfall was greater than 0%. However, participant 14 showed a very consistent pushing style regardless of crossfall gradient, managing to employ coping strategies 1 and 2.

P14 reduced the magnitude of the peak F_t on the upslope side throughout the task, while simultaneously increasing the peak F_t on the downslope side. P14 continued to propel bilaterally (applying contacts to both wheels simultaneously) throughout the task, and only needed to substitute brakes for pushes on the upslope side of the wheelchair in the 4% condition. The pattern of needing to apply a Brake on the 6th Contact was seen on all 3 runs of the upslope runs for P14.

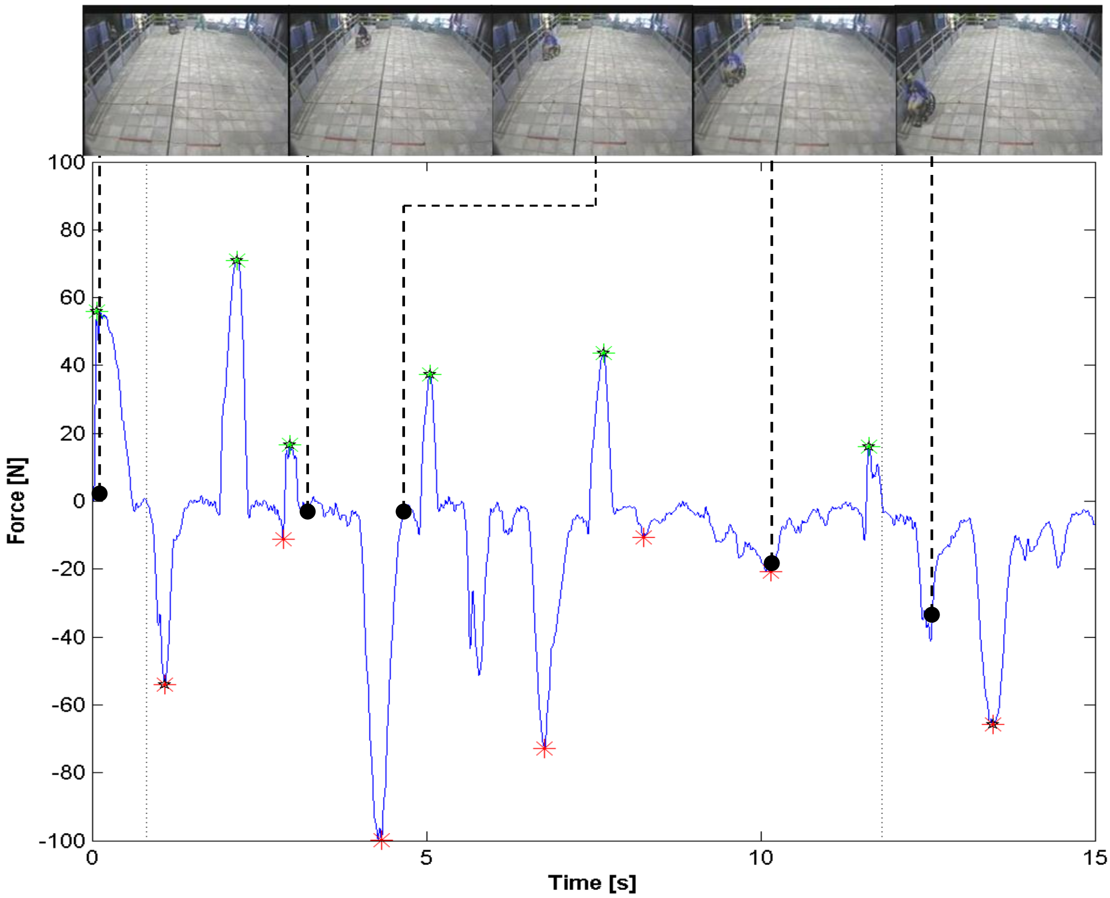


Figure 7-8: Figure showing tangential force for Participant 1 on the upslope side. Illustrated with video snapshots showing deviation from a straight line

The provided capabilities of the SPWS when P1 was an occupant were markedly different to when P14 was the occupant. This is despite both participants being female, of similar age (28 and 33 years respectively) and weight (59.9kg and 60kg respectively).[19] As the occupant mass is the same, the amount of going Work should be the same for equal SPWS velocities. P1 had a MVPF of 104 N and P14 had a MVPF of138 N, both of which should be sufficient for the task. However, the SPWS deviated 0.73m on average on the 2.5% crossfall, and a full 1.2m on the 4% crossfall when P1 was the occupant. This is highlighted using a single trial's data in Figure 7-8. In contrast, when P14 was the occupant there was no noticeable deviation. It does not appear to be the case that the cause of this difference would lie in a lack of strength as the MVPF of P1 is much higher than the peak forces applied by P14 (see 5.2). Therefore, it can be concluded that skill and technique are more influential as has just been seen from comparing the individual Coping Strategies of these participants.

[19] The differences between P1 and P14 have been previously discussed by Holloway *et al* (2010).

7.3.3 Push Patterns Revisited

In section 3.3 it was suggested that with three key parameters the full pushing pattern could be described. These were the peak force, the time of the contact and the number of contacts. The following section of the discussion investigates if it would be useful to assess wheelchair accessibility in terms of these 3 key parameters. To do this 3 'task' triangles, one for each contact type were constructed. These are shown in Figure 7-9 for participant 14 (top row) and participant 1 (bottom row) for the upslope side of the wheelchair. In order to construct the triangles average values over the 3 runs of Peak Ft, Contact Time and Contact number were calculated and the base of each triangle constructed using the mean contact time. The apex represents the mean peak F_t of the contact multiplied by the number of contacts. The green triangles represent the total push force, the orange triangles the total impact force and the red triangle s the total braking force.

Using this visual type of representation of the data may be useful in showing the coping strategy utilised by people. This is shown by the top row, where strategy 3 (reduction in upslope force) is clearly demonstrated in both the decreasing size of the green triangles as crossfall gradient increases, and the increasing size of orange and red triangles. It is also a clear way of representing the difficulty faced by participant 1, especially when compared to participant 14. A further use of this type of graphical representation could be to compare different environmental barriers. In this situation 2 different routes could be easily compared which contain different types of barriers.

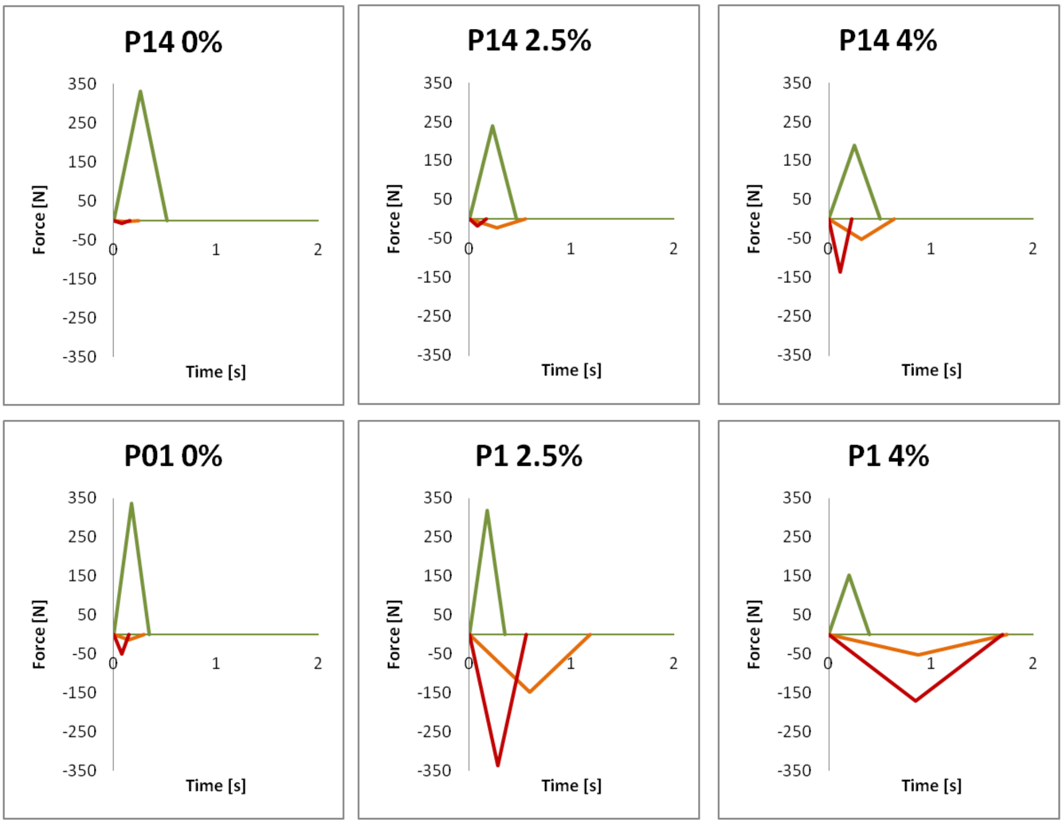


Figure 7-9: Graphical representation of changes in the SPWS provided capabilities

7.3.4 Crossfalls Relative to other Barriers

Starting forces are normally higher than the forces needed to keep the wheelchair moving once started. The mean values found in the current experiment for peak tangential force were 72.1 N, 65.8 N and 69.8 N respectively for the 0 %, 2.5% and 4% crossfall on the downslope side and 63.6 N, 52.4 N and 50.7 N on the upslope side. These values are similar to those reported by Koontz et al (2005) for smooth level concrete (≈82 N ± 22 N)[20].

During the going phase of this study on the Work done on the Upslope and Downslope sides of the wheelchair, the results showed that the increase in work per contact on the downslope side was significant, increasing from a mean of 57 Nm on 0% to 71 Nm on 2.5% and 94 Nm on 4%. These values were compared to those computed in a recent study by Hurd *et al.* (2009). They found that on average (between non-dominant and dominant hand) 13.55 Nm, 18.6 Nm and 24.4 Nm was done per push on smooth concrete, aggregate

[20] Koontz *et* al reported Peak Wheel Torque, which is analogous to the tangential force used in this study as both studies utilise the SmartWheel, and so the handrim diameter are the same. The moments reported by Koontz have been converted here to Forces by dividing by the radius of the SmartWheel handrim.

concrete and a 3 degree (approximately 5.4%) incline (Hurd *et al.* 2009). These values are markedly less than those reported in this thesis.

This was briefly investigated by analysing the work done for a single push using the 4th push of participant 7's (a wheelchair user) run on the 0% condition. The work was calculated firstly by integrating M_z with respect to the angle[21], and then integrating F_t with respect to distance (see Figure 7-10). There is a noticeable increase in work when calculated using the second method, which partly explains the discrepancy between the values reported in this Thesis and those of Hurd *et al.* A second explanatory factor for the discrepancy may the differences of the rolling resistance of the two types of concrete surface tested by Hurd *et al.* and the concrete paver surface used in the current experiments.

The method used in this thesis is preferred as a measure of the Provided Capability of the SPWS as it represents the work done in actually moving the wheelchair along the footway. Conversely, when the moment is integrated with respect to the angle the work being

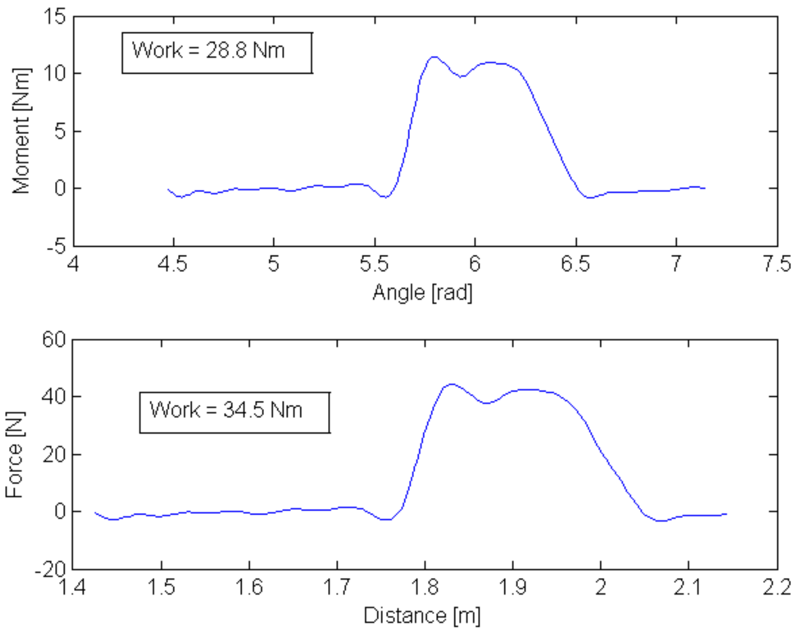


Figure 7-10: Illustration of the two different ways to measure work. Top shows a plot of Wheel Moment, Mz, (Nm) against wheel angle (rad). Bottom shows a plot of Tangential force, Ft, (N) against distance (m). Both plots are annotated with the amount of work as measured by integrating under the curve shown.

calculated is that done in moving the wheel.

[21] The angle was converted from degrees to radians before calculating the work. Mz, Ft and distance were all direct outputs from the SmartWheel's Format 2 files used in this study

7.3.4.1 Impacts and brakes

The prevention of upper-limb injury has been linked to smooth pushes on the handrim. There has been no study conducted into the how applying a braking force to the handrim compares with the application of a push. It could be assumed that although a push and brake are different, the application of a smooth braking force is preferred to a short sharp braking force. Impacts are a little more difficult to classify as they sometimes increase in length and magnitude to such an extent that they are larger than the brakes. The fact that impacts occur immediately before a push may mean that the transition from applying a negative force to a positive force is in fact more detrimental to the upper limb than simply applying a brake then releasing the handrim and applying a push. These questions are raised as points of further discussion as they fall beyond the scope of this thesis.

7.4 Conclusions

Given the results of the experiments into the forces necessary to start and keep a wheelchair going (with a total mass of 104 kg) it can be concluded that manual handling guidelines for practices in the work place are having to be broken by attendants when pushing a wheelchair along a footway with a crossfall gradient of greater than 0%. However, given the small sample size it would be erroneous to recommend changes to either current NHS provision policy, or the footway guidelines. However, the subject of attendant-propulsion of wheelchairs is in need of further research.

The current study utilised a 9L wheelchair, and it is possible that the required capabilities would be reduced if an 8L were used to negotiate the same task, as theoretically the larger rear wheel should reduce the rolling resistance of the wheelchair.

It could be suggested that clinicians keep in mind the maximum push force guidelines when providing a wheelchair which will be pushed predominantly by an attendant, and consider the provision of a lighter wheelchair or a power-pack[22] when the occupant mass is large relative to the attendants strength. This should particularly be borne in mind when the attendant is elderly and female.

[22] A power-pack is small pack which can be attached to the rear of a wheelchair to aid forward motion. It consists of a small motor, which delivers power to a small, additional wheel.

With regards to self-propulsion of wheelchairs on crossfalls it can be concluded that technique is a very important factor in determining how a person is able to cope with a crossfall of greater than 0 %. Upper-limb injury has been linked to short sharp push forces of high magnitude. For this reason it is recommended that occupants firstly reduce the push force they apply to the upslope handrim, and if necessary increase the amount of force on the downslope side. A combination of these two strategies is seen as the best compromise between maintaining the desired velocity level and reducing the risk of upper limb injury. Negative forces are only recommended when there is a risk of injury or accident as they are an inefficient mode of wheelchair propulsion and so utilise the upper limbs without a direct benefit in wheelchair motion.

The Capabilities Model has provided a good framework for measuring the accessibility of footways for wheelchair users and has led to the development of a new visual representation of wheelchair self-propulsion patterns.

8 Further Research

Now that a methodology has been developed for analysing crossfalls, it is necessary to test a greater number of people and to investigate longer tasks than the 10.2m footway tested on this occasion. It could well be the case that a limiting factor in terms of user ability to propel a wheelchair is their aerobic fitness rather than their strength and so this should also be investigated.

Other complex terrains such as road crossing s and avoidance of obstacles should be investigated. The results of experiments using a variety of different terrains could then be combined to classify routes as easy or hard. This could be done using the triangles for self-propulsion with analogous rectangles for attendant propulsion.

Specific areas of further work for attendant propulsion (Section 8.1), self-propulsion (Section 8.2) and the capability model (Section 8.3) will now be described.

8.1 Attendant-propulsion

The following topics are suggested for further research within the area of attendant-propulsion:

1. The difference between the *Provided Capabilities* used when pushing an 8L compared with a 9L wheelchair. As these are both standard issue wheelchairs from the NHS, it would be useful to know if the 8L is able to deliver a reduction in the *Provided Capabilities* of the APWS and reduce the burden on the attendant. Furthermore an investigation into the effect of handle height and other component changes to the *Provided Capabilities* should be investigated.
2. The effect of the gait cycle parameters on push forces of attendant propulsion.

8.2 Self-propulsion

The following topics are suggested for further research within the area of self- propulsion:

1. Upper body posture when traversing crossfalls. If people adjust their body posture so that they are sitting upright in the wheelchair this will make them further away from one handrim and nearer the other. This could be one of the factors which affect the pushing patterns of people as on the upslope side they might have a

longer stroke and on downslope side a shorter stroke. It is also possible that people reduce the amount they move their trunk forward at the end of a push when on a crossfall greater than 0 %. The reason could be that the added acceleration may only serve to accelerate the wheelchair downslope if the forces are slightly misaligned in terms of the force difference necessary to overcome the crossfall.

2. The effect of wheelchair set-up on the *Provided Capabilities* of the SPWS. In general if a user has the ability to flip their casters and thus perform a wheelie, their wheelchair will be set-up to allow them to do so. This involves moving the rear axle of the wheelchair forward, which in turn reduces the distance between the centre of mass of the SPWS and the rear axle. This, theoretically, should reduce the tendency of the wheelchair to turn downslope.

8.3 Capability Model

The following could be done to further develop the *Capabilities Model* for assessing wheelchair accessibility. Firstly, it would be useful to measure a cohort's *Provided Capabilities* that are considered necessary for self-propulsion of wheelchairs and attendant-propulsion of wheelchairs. These could then be compared to the *Required Capabilities* as measured by such physical forces such as rolling resistance. This would allow a fuller picture of the Capabilities to be gleaned and possibly lead to the ability of the model to predict thresholds of *required capabilities*, which could be used in future, built environment guidelines.

The idea of visual representations of the impact of barriers on wheelchair users, shown by the 'triangle' graphs in section 7.3.3 could be further developed to compare across different barriers.

9 Conclusions

It was shown in Chapter 2 that the capabilities model was able to represent the various interactions that occur when a wheelchair user wants to traverse a footway. Within this framework, it was shown that there exist a number of required capabilities which must be overcome in order for a wheelchair to move along a footway. Further, it was indicated that in order for a wheelchair and its user(s) to move from A to B they would need to employ a number of provided capabilities, which would need to be greater than the required in order to be able to achieve their aim. Chapter 2 concluded that in order to push along a standard UK footway a wheelchair user would have to be able to counter the required capabilities of a 2.5% crossfall gradient, assuming the footway was within the current guidelines.

Chapter 2 also explored previous literature with regards injuries to wheelchair users and the links between injuries and force measurements. There was shown to be a link between the type of force applied to the handrim of a wheelchair by an occupant and injury levels. There was also shown to be a link between pushing in general and back and shoulder injuries. The fact strength decreases with age was also shown.

Chapter 2 finally concluded there had been differences in approach and conclusions made by previous investigations into the effect of crossfall gradient on wheelchair propulsion.

In Chapters 3 and 4 the methodologies for self-propulsion and attendant-propulsion respectively were laid out to test the overarching hypothesis of this thesis. The hypothesis was that in order to overcome a crossfall of greater than 0% users would need to utilise an increased number of provided capabilities. It was thought that in order to move the wheelchair along the footway a person must possess the provided capabilities necessary to start, keep going and stop the wheelchair.

The provided capabilities need to start and stop the wheelchair were represented by the peak force necessary to accomplish each task. The going phase was represented by the sum of work done to the wheelchair on the upslope and downslope sides and the difference of work done on the upslope and downslope sides.

The results of Chapter 5 (self-propulsion) and Chapter 6 (attendant-propulsion) concluded that while the sum of work provided to keep the wheelchair moving along a footway was independent of crossfall gradient, the difference of force increased with crossfall gradient.

This showed that an increased number of capabilities were provided by the users in order to overcome the effect of the crossfall. The difference in work was made up with a combination of a reduction in upslope work and an increase in downslope work. For self-propulsion downslope work was reduced by the introduction of increasing negative work in the form of impacts and brakes. This was combined with a decrease in positive work produced by the pushes. Occupants also increased the amount of downslope positive work by increasing the amount of work done by the pushes. Attendant propulsion differed slightly in that there was very little negative work applied to either handle; this meant the difference in force was achieved by reducing the upslope work and increasing the downslope work.

Chapter 5 showed that one wheelchair user (Participant 1) was unable to complete the task when the crossfall was greater than 0 %, and she deviated the full 1.2m of the footway for all runs when the footway gradient was 4 %. However, the MVPF result for this participant showed she had produced sufficient strength to complete the task. All attendants could complete the task, proving they each had the provided capabilities necessary.

The forces necessary to start and stop a wheelchair when pushed by an attendant were reported in Chapter 6 and compared to manual handling guidelines in chapter 7. It can be concluded that the forces provided by some of the attendants tested to start and keep a wheelchair going (with a total mass of 104 kg) broke the guidelines when the crossfall was greater than 0%. It was concluded that this area was in need of further research.

The pushing patterns used by occupants to propel the wheelchair were reported in Chapter 6 and these were further analysed in Chapter 7, where a new visual aid was proposed to categorise barriers in the built environment for wheelchair users. This aid helps to visualise the provided capabilities used by the occupant in response to the required capabilities imposed on them. It was concluded that development of this aid would be worthy of further research.

In summary, it can be concluded that the way in which the environment interacts with the wheelchair-occupant-attendant system can be characterised by the relationship between the required and provided capabilities in each case. Within the bounds of this interaction, the current guidelines for footways with regard crossfall gradient appear adequate for short distances (i.e. < 10 m), given that the only participant to not have the provided capabilities was not a regular wheelchair user. The methodology suggested in this thesis could be used to establish the required and provided capabilities pertaining to longer distances and more complex environments. An important outcome is that, in addition to the increase in capabilities required to travel a distance (e.g. strength), the introduction of a crossfall gradient of greater than 0 % requires the user to utilise an additional capability – the ability to provide differential forces simultaneously on both sides of the wheelchair – and it may be the case that some users do not possess this capability. Finally, it can be concluded there is clear scope for further research in this area, and details of possible future topics have been given in Chapter 7.

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Appendix 1: Conversion Table of Footway Gradients

Gradient [%]	Gradient [degrees]	Gradient [radians]
2.5	1.43	0.02
4	2.29	0.04

Appendix 2: Terms used in this thesis

Attendant: someone who pushes a wheelchair using the rearward facing handles and walking behind the wheelchair
Attendant-propulsion: The propulsion of a wheelchair by an attendant using the rearward facing handles, whilst walking behind the wheelchair. Otherwise referred to as a 'transit' wheelchair.
Brake(s): Negative contact with the handrim of a wheelchair that takes place in lieu of a push.
Capabilities Model: A model describing the interactions between what the capabilities available to a person (provided capabilities) and those needed to complete a task (required capabilities)
Coping Strategy: the method employed by someone to overcome a barrier
Crossfall: the lateral slope on a footway that aids surface water drainage
Downslope: The side of the wheelchair which is higher than the other when traversing a footway with a crossfall gradient of greater than 0%. When the crossfall gradient is equal to 0% the downslope side was arbitrarily defined as the runs which went South.
Downslope negative work: The work done as a result brakes and impacts applied to the handrim on the downslope side of the wheelchair
Downslope positive work: The work done as a result pushes applied to the handrim on the downslope side of the wheelchair
Downslope Work: The work done as a result of force applied to the downslope side of the wheelchair
F_{tPK}: The peak tangential force applied to the handrim during a contact with the handrim
Going Phase: the going phase consists of the period of time between starting and stopping a wheelchair. For attendant propulsion this is defined as From the time of the first local minimum found in Ftot when it is plotted against time to the time a negative (braking) Ftot is applied to the handles. For self-propulsion it is defined as starting

when the occupant makes contact with the handrim for a second time and ends after the last positive contact.
Going Work: The amount of work necessary to move the wheelchair along the footway during the going phase.
Impacts: Negative contacts with the handrim that occur immediately before or after a push
Matching Runs: participants initially went along the footway in one direction and then came back in the opposite direction. These runs were 'matched' in order to calculate the sum and difference of forces and work done on the wheelchair for each condition. Therefore, 'upslope' and 'downslope' runs were matched by trial number, crossfall gradient and participant number.
Maximum Voluntary Push Force: An isometric test of push force where the wheelchair is restrained and the user asked to push as hard as they can on the handles in the case of attendant propulsion and on the handrim in the case of self-propulsion
Negative Downslope Work: The work done on the downslope sides by brakes and impacts
Negative Upslope Work: The work done on the upslope sides by brakes and impacts
Occupant: the user who sits in the wheelchair
Peak Starting Force: The peak force needed to start the wheelchair. In the case of the attendant this is the peak force (left and right handle forces combined) necessary to start the wheelchair moving at a particular speed. For self-propulsion it is the value of the peak of the first push force
Peak Stopping Force: The peak stopping force needed to stop the wheelchair.
Peak Stopping Force: The peak value of the tangential force during the Stopping phase.
Positive Downslope Work: The work done by the force applied to the wheelchair on the downslope side by the pushes/positive force in the case of self-propulsion. In the case of attendant-propulsion it is the positive force applied to the downslope handle.

Positive Upslope Work: The work done by the force applied to the wheelchair on the upslope side by the pushes/positive force in the case of self-propulsion. In the case of attendant-propulsion it is the positive force applied to the upslope handle.
provided capabilities: The abilities person posses which can be utilised to achieve a task.
Provided Going Work Difference: The difference of the Going Work from upslope and ‘matching’ downslope runs, where the value from the upslope run is subtracted from the value obtained in the downslope run.
Provided Going Work: The sum of the Going Work from the upslope run and the Going Work from the ‘matching’ downslope run.
Provided Starting Force: The sum of the Peak Starting Force from the upslope run and the Peak Starting Force from the ‘matching’ downslope run.
Provided Stopping Force Difference: The difference of the Peak Stopping Force from upslope and ‘matching’ downslope runs, where the value from the upslope run is subtracted from the value obtained in the downslope run
Provided Stopping Force: The sum of the Peak Stopping Force from the upslope run and the Peak Starting Force from the ‘matching’ downslope run.
Push(es): (A) positive tangential push force applied to the hand rim.
Quasi-Steady-state Phase: The time period for which the velocity was thought to be at its most constant. This was found by visual inspection of velocity against time plots.
required capabilities: The capabilities needed to complete a task.
Rolling Resistance Force: The friction between the wheelchair and the paver surface, which must be overcome when the wheelchair is travelling at a constant velocity. It is assumed that air resistance, bearing friction, and energy losses within the frame are minimal and are therefore ignored.
Self-propulsion: The propulsion of a wheelchair by the occupant using the rear-wheel handrim(s) of the wheelchair
SmartWheel: A commercially available wheel, which is capable of measuring 3-

dimentional forces and moments as well as velocity. It can be attached to a wheelchair in place of the standard wheel to measure pushrim kinematics.
Starting Force(s): The peak force applied to the wheelchair during the starting phase
Starting Phase: The phase which involves the acceleration of the wheelchair from rest to the going phase
Stopping Force(s): The peak force applied to the wheelchair during the stopping phase
Stopping Phase: From the time a negative (braking) F_{tot} is applied to the handles to the time the wheelchair stops moving.
Total Horizontal Force: The sum of the horizontal components of the force transducer readings from the right and left handles.
Upslope: The side of the wheelchair which is higher than the other when traversing a footway with a crossfall gradient of greater than 0%. When the crossfall gradient is equal to 0% the upslope side was arbitrarily defined as the runs which went North.
Upslope negative work: The work done as a result brakes and impacts applied to the handrim on the upslope side of the wheelchair in the case of self-propulsion. In the case of attendant-propulsion it is the negative force applied to the upslope handle.
Upslope positive work: The work done by pushes on the upslope side of the wheelchair in the case of self-propulsion. In the case of attendant-propulsion it is the positive force applied to the upslope handle.
Upslope Work: The work done by the forces applied to the upslope side of the wheelchair
User: either attendant or occupant of a wheelchair
Wheelchair system: refers to the combination of the occupant and the wheelchair

Appendix 3: Measuring the required capabilities when propelling a wheelchair along a footway

Definitions

Required Capabilities (C_{RQD}): The capabilities needed to complete a task.

Total Horizontal Force (F_{tot}): The sum of the horizontal components of the force transducer readings from the right and left handles.

Start-up Phase: From the a positive F_{tot} is applied to the handles to the first local minimum found in F_{tot} when it is plotted against time (see).

going Phase: From the time of the first local minimum found in F_{tot} when it is plotted against time (see) to the time a negative (braking) F_{tot} is applied to the handles.

stopping Phase: From the time a negative (braking) F_{tot} is applied to the handles to the time the wheelchair stops moving.

Quasi-Steady-state Phase: The time period for which the velocity was thought to be at its most constant. This was found by visual inspection of velocity against time plots.

going Work (CR_{wk}): The amount of work necessary to move the wheelchair along the footway during the going phase. This is defined as the time from when the force horizontal force is greater than zero at the start of the experiment, to the move the wheelchair at any given velocity.

Peak Starting Force (CR_{start}): The peak F_{tot} necessary to start the wheelchair moving at a particular speed.

Peak stopping Force (**CR_{stop}):** The peak F_{tot} necessary to stop the wheelchair moving at a particular speed.

Rolling Resistance Force (CR_{RR}): The friction between the wheelchair and the paver surface, which must be overcome when the wheelchair is travelling at a constant velocity. It is assumed that air resistance, bearing friction, and energy losses within the frame are minimal and are therefore ignored.

Required Capabilities: There are 4 Required Capabilities defined. The first is the Peak Starting force (**CR_{start}**), the second the going Work (**CR_{wk}**), the third the Peak stopping Force (**$CR_{stop,}$**) and finally the Rolling Resistance Force (**CR_{RR}**).

Aims & Hypotheses

There are three aims of this section of the thesis. The first is to assess the effect of crossfall gradient on the capabilities required for a wheelchair to start, continue to move and stop on a footway. The second is to assess the effect of velocity on these capabilities and the third is to attempt to measure the rolling resistance of the footways.

Hypothesis 1 Effect of Crossfall gradient

H0: There will be no change in the required capabilities regardless of footway crossfall gradient.

H1: There will be a significant proportional linear relationship between the required capabilities and the footway crossfall gradient.

Hypothesis 2 Effect of Velocity

H0: There will be no change in the required capabilities regardless of the velocity of the wheelchair.

H1: There will be a significant proportional linear relationship between each of the in the required capabilities and the of the velocity of the wheelchair

Theory: Weight Distribution & Downward Turning Moment

In order to locate the centre of mass of each wheelchair system in the plane made by the rear axle and the direction of travel (perpendicular to the rear axle); each wheelchair system was placed on four postal scales and the mass recorded by each scale was noted. The distance between the rear-wheels, the distance between the castors and the wheelbase of the wheelchair were also recorded (see Figure A3-1). These were then used as inputs to Equation 18 and Equation 19.

$$CoM_x = \frac{(m_{rc} + m_{lc}) * wb * 9.81}{m}$$

Equation 18: Equation for the location of the centre of mass of the wheelchair along the x-axis from the left rear wheel. m_{rc} and m_{lc} are the masses recorded under the right and left castors respectively, wb is the wheelbase of the wheelchair and m is the mass of the wheelchair system.

$$CoM_y = \frac{(m_{rc} * c + m_{rw} * d) * 9.81}{m}$$

Equation 19: Equation for the location of the centre of mass along the y-axis from left rear wheel. m_{rc} and m_{rw} are the masses recorded under the right castor and right wheel respectively, d is the perpendicular distance between the rear wheels, c is the perpendicular distance from the right castor to the left wheel and m is the mass of the wheelchair system.

The centre of mass location in the forward direction of travel, and the mass of the system were used as inputs into Equation 20, which calculates the downward turning moment for

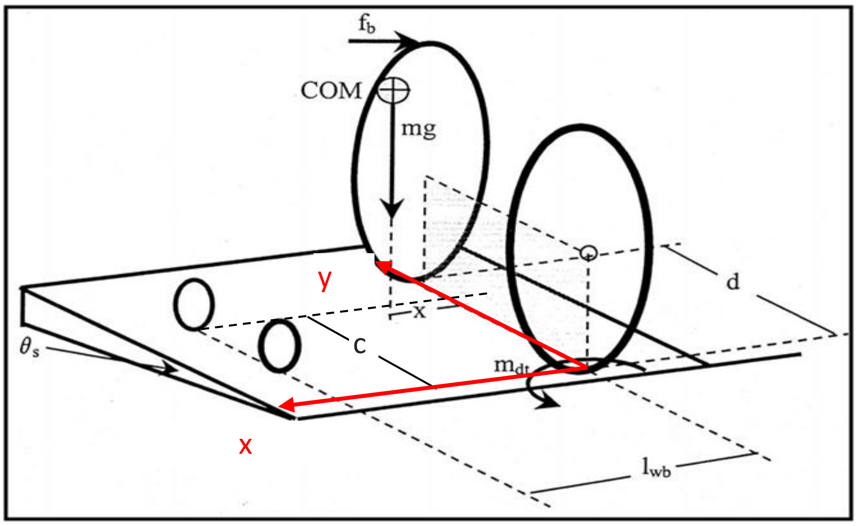

Figure A3-1: Diagram showing the distances and dimensions needed to locate the centre of mass of the wheelchair and which influence the downward turning tendency of a wheelchair on a cross fall. The image has been taken and adapted from Tomlinson (2000). COM is the location of the centre of mass, c is the distance from the right castor to the left wheel, f_b the braking force. The x and y axis are shown in red.

various crossfall gradients (θ_s) (Brubaker *et al.* 1986; Tomlinson 2000b).

$$M_d = Wxsin(\theta_s)$$

Equation 20: Equation to calculate the downward turning moment M_d from the weight of the wheelchair system (W), the perpendicular distance from the rear axle position and the location of the centre of mass in the direction of travel (x) and the gradient of the crossfall θ_s.

The downward turning moment can be overcome by applying a breaking force to the upslope handle or wheel. The resulting braking force necessary to overcome the downward turning moment is then found by dividing the moment by the distance between the 2 rear wheels using Equation 21 (Brubaker *et al.* 1986; Tomlinson 2000b).

$$F_b = \frac{mgxsin(\theta_s)}{d}$$

Equation 21: Equation to calculate the braking force F_b required to prevent a wheelchair turning downslope. The wheelchair system (W), the perpendicular distance from the rear axle position and the location of the centre of mass in the direction of travel (x) and the gradient of the crossfall θ_s.

From the theory above, it is clear there is a downward turning tendency of a wheelchair as it travels along a crossfall. If the castors are prevented from turning downslope, and thus the wheelchair travels in a straight line, it is thought the force necessary to overcome the downward turning tendency of the wheelchair would be overcome by an increase in F_{tot}. This in turn would mean the required capabilities necessary to start (**CR_{start}**), keep going (**CR_{wk}**), and stop (**CR_{stop}**,) the wheelchair would need to increase as crossfall gradient increased. This would be because the rolling resistance would have in effect (**CR_{RR}**), see Hypothesis 1.

Required Capability Testing Methods (Attendant)

The instrumentation of the wheelchair is described in section 0, followed by details of the procedure in section 0.

Wheelchair and Instrumentation

The wheelchair used in this experiment was a standard issue NHS attendant-propelled wheelchair, the 9L (see Figure A3-2). The wheelchair has a wheelbase (distance between rear axle and castor axle) of 36 cm when the castors are trailing back (as shown in Figure A3-2). The rear wheel track (distance between the rear wheels) is 50 cm.

The wheelchair was instrumented so that the handle forces and also the rear wheel speeds could be recorded. The handle forces were recorded by attaching two 6-axis force transducers in line with the rubber grips of the handles. The force transducers used are commercially available and produced by AMTI (model MC3A-6-250). The force signals were amplified using amplification boxes again from AMTI (model MSA-6). The force data from the direction of travel (F_x) from both handles was recorded using a datalogger (Measurement Computing, model USB-5201). Both rear wheels had a rotary encoder to detect the velocity. The encoder consisted of a teethed gear which rotated with the rotating rear wheel. The encoder outputs 500 pulses per revolution and this signal was processed using custom circuitry. The resulting voltage was output to the same datalogger as the one used to collect the force data. The accompanying datalogger software (TracerDAQ) was used to record the data files, which was run on a laptop secured to the wheelchair. All data was recorded at a sampling frequency of 100 Hz.

Two metal bars were screwed together above and below the instrumented wheelchair handles (see Figure A3-2). A rope was attached between the centre of the 2 bars joining the handles and a metal bar attached to the electric scooter (see Figure A3-2). The rope was attached at the same height as the handles so that it would remain as close to horizontal as possible.

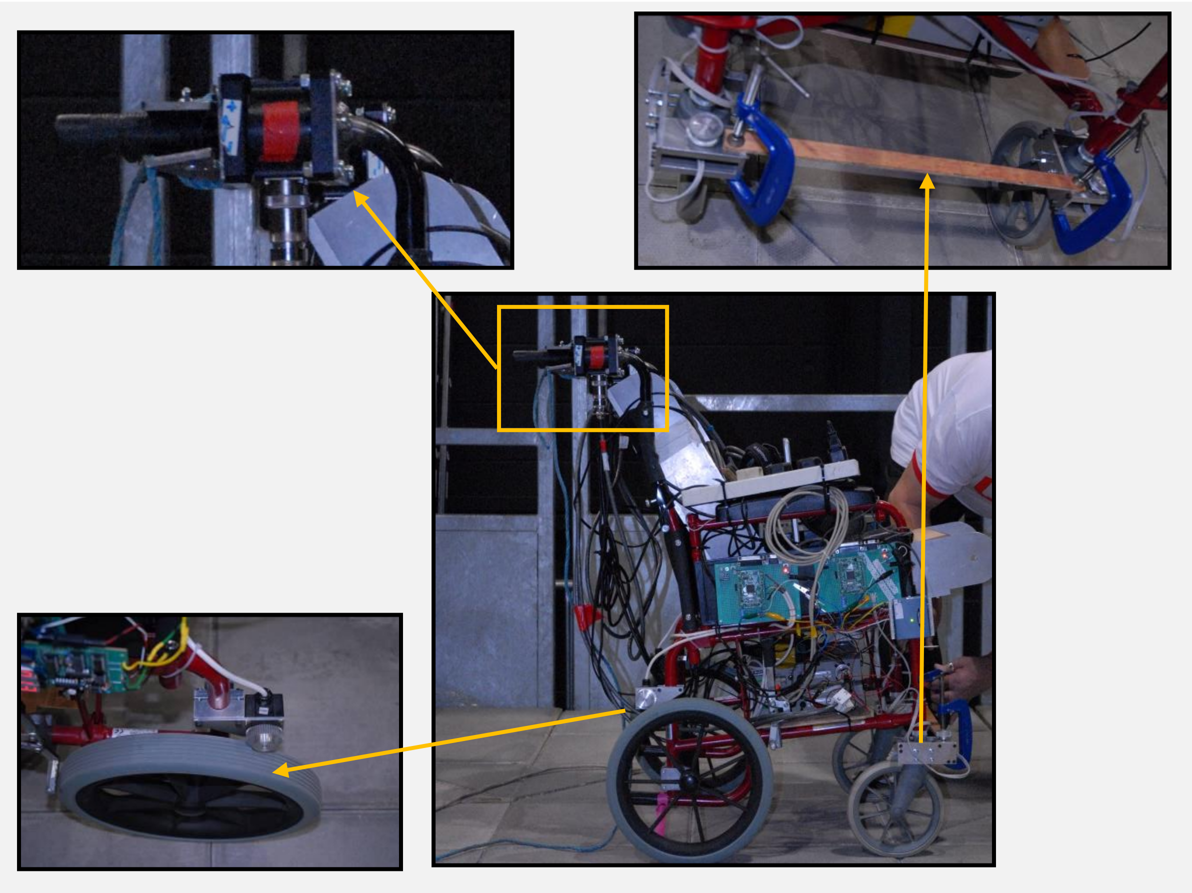

Figure A3-2: Photos of instrumented wheelchair (bottom right) , with details of the rear wheel rotary encoder (bottom left), the right handle force transducer (top left) and the clamping of the front castors (top right).

Experimental Methodology

The wheelchair was weighed by placing it on 4 postal scales, so that one was under each wheel and castor. ISO dummy weights were added to the wheelchairs to simulate an occupant mass of 75 Kg. The locations of the weights were adjusted to ensure the weight was distributed evenly between the left and right sides of the wheelchair.

The wheelchair was pulled along 3 footways of different crossfall gradient: 0%, 2.5% and 4%. Each lane was 10.2 m long and 2.4 m wide. Due to being pulled by the scooter, there was an effective length of footway of 7.2 m for the wheelchair to travel along (see Figure A3-3).

For each of the three gradients of footway the wheelchair was pulled at 5 target velocities: 0.70m/s, 0.80m/s, 1.00m/s, 1.20m/s and 1.45m/s. These velocities were chosen as they corresponded to easy-to-locate positions on the scooter's velocity dial. Each velocity was repeated at least once.

The wheelchair was lined up behind the start line, and was pulled by the scooter at the desired speed. On occasions the wheelchair drifted noticeably from the straight line, and these trials were not saved, and the experiment was repeated. The wheelchair was stopped by an attendant pulling on the handles.

Figure A3-3: Experimental Procedure of pulling the wheelchair with the scooter. The wheelchair is attached to the scooter via a rope connected between the metal bars connecting the handles and a metal bar attached to the rear of the scooter.

The data were recorded as described in section 0. The files were imported into Matlab (Version 7.10.0.499, R2010a) and analysed using a custom script.

The Matlab script first filtered all data channels with an 8th order low pass Butterworth filter with a 10Hz cut-off frequency, to remove the effect of vibrations. The Total push Force (F_{tot}) was found by summing the right and left handle forces and the Average Velocity (V_{avg}) was found by calculating the average velocity of the right and left wheels.

Figure A3-4 shows an example force (top) and velocity (bottom) plots against time. Also shown are the start and end times of the 'going' and 'QSS' phases. The start time of the going phase was found by using the peakdetect function to find the local minimum value of F_{tot} following the first local maximum (start-up peak). The end time of the going phase was found by stepping back through F_{tot} from the last local minimum (stopping peak) until force was greater than zero. The start time of the QSS phase was determined by visually inspecting each run by plotting the velocity curves against time (as in Figure A3-4). The start time was chosen so that the resulting velocity was at its most constant (i.e. there was no acceleration).

The peak start push force was then found using the built-in max function and the peak force

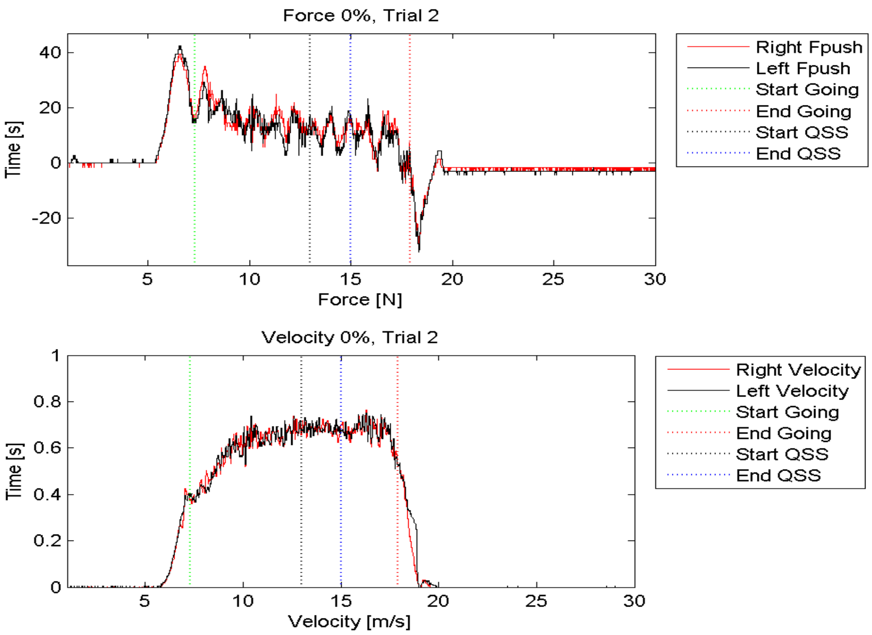

Figure A3-4: Sample force (top) and velocity (bottom) traces showing the start and end points of the Quasi-Steady-State and Going phases

used to stop the wheelchair found using the built-in min function. The Average Velocity was integrated with respect to time to find the distance travelled. Ftot was then integrated with respect to distance to calculate the work in Nm. The average force and velocity of both wheels for the quasi-steady-state (QSS) phase was also calculated. The force required to keep going in the QSS phase is equal and opposite to the Rolling Resistance provided the wheelchair system is not accelerating at the time.

Each file was checked to ensure it did not deviate from a straight line by plotting the left wheel velocity against the right wheel velocity. An example curve is given in Figure A3-5. No curve appeared to have a systematic deviation from a straight line through the origin. However, there was a degree of scatter as shown in Figure A3-5.

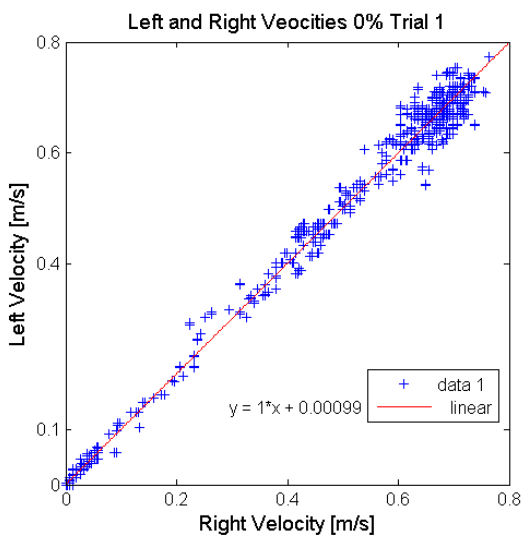

Figure A3-5: Example plot of left wheel velocity against right wheel velocity, showing the linear relationship between the two velocities.

Required Capabilities Results

Weight distribution of wheelchair

Each wheelchair system was weighed by placing it on 4 postal scales and recording the readings from the scales as described in section 0. The results of the weights (in kilograms) recorded for each contact point between the wheelchairs and the 4 postal scales are shown in Table A3-1.

Contact point	Attendant Mass (kg)
Right Wheel	33.70
Left Wheel	35.15
Right Castor	21.15
Left Castor	21.25
Total Mass	111.25

Table A3-1: Recorded weights under each wheel and castor

The total mass of the attendant wheelchair system was 111.25kg, which was distributed with a near 50:50 balance between left and right sides and almost a 60:40 balance between rear and front (see Table A3-2 for exact values). The total mass of the self-propelled wheelchair system was 93.8 kg. This was also distributed evenly between left and right sides (see Table A3-2). The weight was distributed with less weight over the rear wheels compared to the attendant-propelled wheelchair, with a 75:25 balance approximately (see Table A3-2 for exact values).

Using Equation 18 the centre of mass for the attendant wheelchair was calculated as being 0.14 m forward of the rear axle for the attendant-propelled wheelchair and. Therefore, on a 2.5% crossfall there is a static downward turning moment of 3.74 Nm acting on the wheelchair; and on a 4% crossfall the moment is 5.98 Nm. Theoretically, this can be overcome by applying 7.6 N and 12.21 N respectively of force to the upslope handle.

However, during these experiments the participants did not need to hold the wheelchair when it was stationary to prevent the wheelchair turning downslope.

Side of wheelchair	Attendant % of total mass
Right	49.39
Left	50.61
Front	38.11
Back	61.89

Table A3-0-2: Mass distribution of wheelchair system

Required Capabilities for Attendant-Propelled Wheelchair

The results of the 4 required capabilities are now reported. Despite fixing the castors to help prevent the wheelchair turning downslope, there were occasions where the wheelchair still travelled off-course and downhill. Although these trials were deleted and not analysed, the fact the wheelchair still managed to travel off-course is note-worthy. This probably occurred due to the wheels and castors slipping sideways down the crossfall.

Quasi-Steady-State & Rolling Resistance

The effects of velocity and crossfall gradient on the push force required for wheelchair propulsion in a straight line during the 'QSS' phase along a footway with a 0% , 2.5% and 4% crossfall gradient are shown in Figure A3-6. It shows a general trend of increased force with crossfall gradient, and a less pronounced increase of force with increasing speed across all crossfall gradients.

As there is assumed to be no acceleration acting on the wheelchair during the QSS phase, the rolling resistance can be estimated by getting a line of best-fit through the points for any given crossfall condition. This is made somewhat difficult especially for the 0% condition due to there being an unusually high and low value in the data. These points are highlighted in Figure A3-6, along with a high value found for the 4% condition. The spread in the data is probably due to the rough concrete surface and the chamfers at the edges of the pavers, which make it difficult to collect smooth force data.

The mean values for each crossfall and each velocity are given in Table A3-3. Despite the variation in the data (see Figure A3-6), the mean values for each target velocity increase

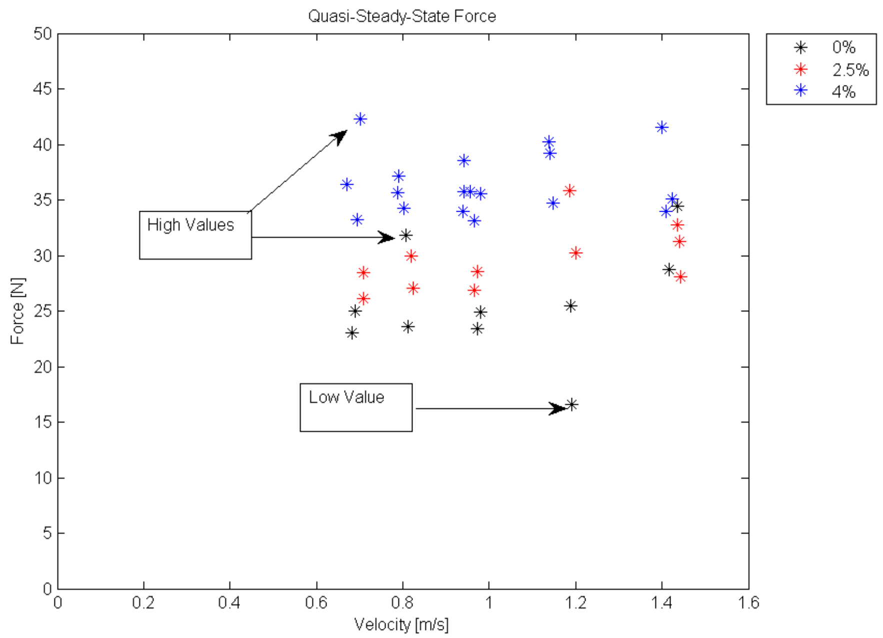

Figure A3-6: Figure plotting the average force during the Quasi-Steady-State Phase, showing a general trend of increasing force with increasing velocity and highlighting the unexpected high and low values.

with crossfall gradient. This is evident when one looks across the rows of Table A3-3. The relationship between the target velocity and the force is less clear and it would appear that given the limited range of velocities tested in this study it is not possible to find a clear linear relationship between velocity and the QSS force. These general trends are backed up by the results of a multiple linear regression model, with crossfall gradient and target velocity as regressor terms. It should be born in mind that the sample size is small for a multiple regression analysis. However, the results of the analysis do produce a reasonable fit to the data (R_{adj}^2=.704, F(2,37)=47.302, p<.001). The constant term and the β-coefficient for crossfall gradient are both significant (p<.001). However, the coefficient for target velocity is not significant (p=.645). Thus, we can conclude that there is a significant effect of crossfall gradient on the QSS force, and therefore the rolling resistance, and that this increases approximately 3.4N with each percentage increase of crossfall. The null hypothesis regarding crossfall gradient can be rejected. However, the null hypothesis regarding the effect of velocity on cannot be refuted.

Target Velocity [m/s]	0% F_{SS} [N]	2.5% F_{SS} [N]	4% F_{SS} [N]
0.70	24.01	27.27	37.29
0.81	27.68	28.51	35.69
1.00	24.17	27.69	35.48
1.20	21.00	33.06	38.07
1.45	16.80	30.68	36.87

Table A3-3: Mean values of force done in the going Phase for each target velocity.

<u>Going Work</u>

There is a general trend of an increase in going Work (CR_{wk}) with increasing crossfall gradient (see Figure A3-7 for individually plotted values and Table A3-4 for the mean values). This trend would be expected following on from the proven increase in force required with crossfall gradient. There also appears to be a clear difference in Work required when the crossfall is greater than 0%.

There are, however, three unexpected low values in CR_{wk}, which are circled in Figure A3-7. All 3 occur on the crossfall and are likely due to the wheelchair not travelling in a completely straight line, despite visibly doing so.

Target Velocity [m/s]	0% Wk_{going} [Nm]	2.5% Wk_{going} [Nm]	4% Wk_{going} [Nm]
0.70	181.58	199.02	229.64
0.81	206.85	226.06	235.27
1.00	207.04	248.28	250.60
1.20	207.24	274.03	282.27
1.45	249.19	237.57	263.68

Table A3-4: Mean values of Work done in the going Phase for each target velocity.

A multiple linear analysis was carried out on CR_{wk} with crossfall gradient and target velocity as regressor terms to see if the trends were statistically significant. The resulting model was a reasonable fit; explaining 45% of the variation seen ($F(2,37)=17.262$, $p<.001$). The coefficients and their significance levels are shown in Table A3-5. It can be seen that there is a significant increase in going Work as crossfall increases, and also as velocity increases

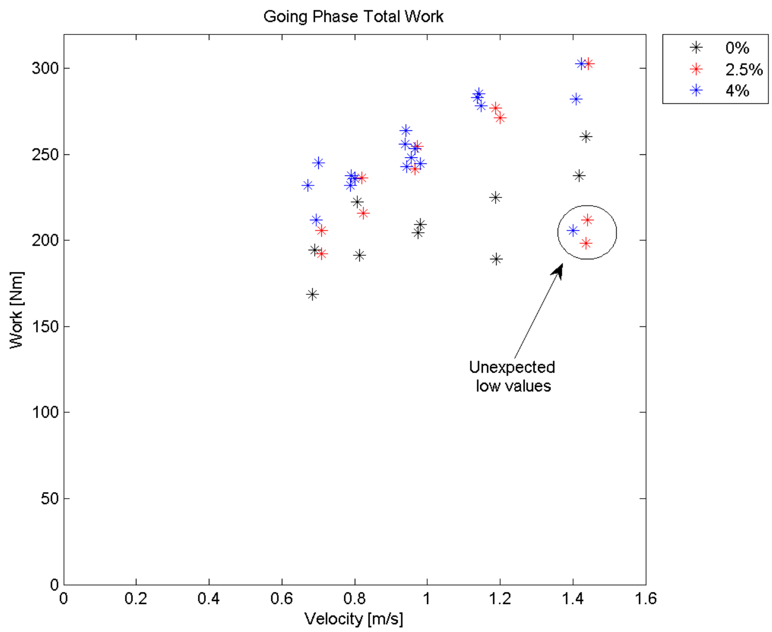

Figure A3-7: Going Phase Total Work showing a general trend of increased work with velocity and with crossfall gradient.

($p<.001$). The constant value is also significant ($p<.001$). A 1m/s increase in velocity would require an additional 58.52 Nm of work, while a 1% increase in crossfall gradient would necessitate an extra 10.32 Nm of work.

Table A3-5 also gives the coefficients for the QSS model. Comparing the rows, the constant terms appear consistent with each other: 23.8 N applied over 6.5 m (the approximate 'going' phase distance in this experiment) would be 154.7 Nm, which is similar to the value found for going Work (151.47 Nm). The values for crossfall are less similar, with crossfall gradient accounting for an increase in force of 3.42 N per percentage increase in crossfall, while traversing approximately 6.5 m would only require an additional 10.32 Nm of going Work according to the models. However, given the relatively small sample sizes they are not so disparate.

The fact that the effect of velocity is found to be significant for going Work and not for QSS force, is probably due to the rather short period of time over which the QSS force was calculated and the erratic nature of the forces over this time period (see Figure A3-4).

Dependant Variables	Model		Coefficients					
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P	Target Vel. [m/s]	P
QSS Force	.719 (.704)	<.0001	23.80	<.0001	3.42	<.0001	-1.03	.645
going Work	.483 (.455)	<.0001	151.47	<.0001	10.32	<.0001	58.52	<.0001

Table A3-5: A summary of the multiple regression analysis for QSS Force and going Work.

In conclusion, the null hypothesis for both the effect of crossfall gradient and velocity can be rejected for going Work as both have a significant effect. A 1% increase in crossfall gradient is approximately equivalent to a .17m/s increase in velocity.

Peak Starting & stopping Forces

The peak starting and stopping forces are, as one would expect, higher in magnitude to the QSS Force. However, they both follow the same general trend of increasing their absolute value with velocity and crossfall gradient (see Figure A3-9. general trend of increasing peak force with velocity for each of the 3 crossfall gradients can be seen in Figure A3-9 and Figure A3-10.

In Figure A3-9 the increase in Starting Force is clear for the 4% condition (apart from 2 unusually high values for the 0%), compared with the 0% and 2.5% conditions. There also appears to be an increase, albeit less marked, with velocity on Peak Start force.

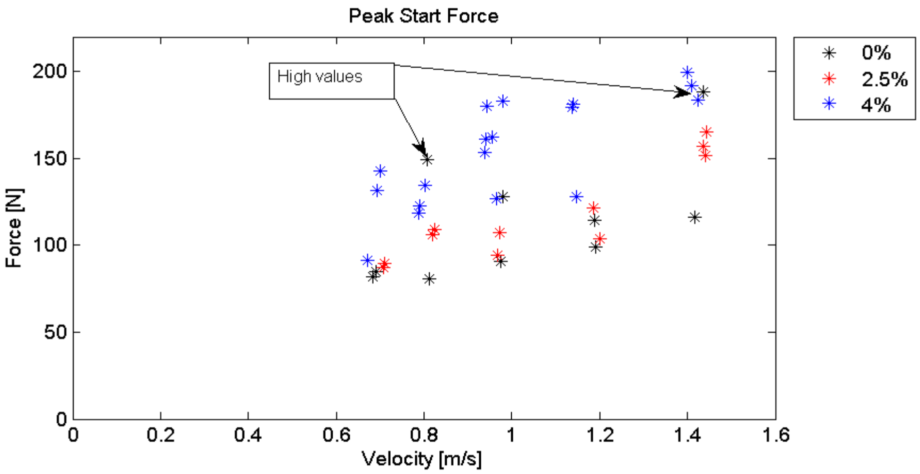

Figure A3-8: individual peak Starting Forces against velocity for each of the crossfall gradients, showing a general trend of increased Starting Force with crossfall gradient and velocity.

The results of the linear regression for Peak Starting Force found a significant increase in Peak Starting Force with crossfall gradient and velocity (p<.001). Overall the model was a reasonable fit, accounting for approximately 60% of the variation seen (R_{adj}^2=.592, F(2,37)=29.307, p<.001). A summary of the model is given in Table A3-6.

Dependant Variables	Model		Coefficients					
	R^2 (R_{adj}^2)	P	Const.	p	Cross-fall [%]	P	Target Vel. [m/s]	P
Starting Force [N]	.613 (.592)	<.001	23.12	.158	11.61	<.001	80.03	<.001

Table A3-6: A summary of the multiple regression analysis for Starting Force

A regression analysis was not carried out for the stopping Force as it was somewhat erratic and resulted in the data breaking the assumption of homoscedasticity due to the larger spread in stopping Forces at higher velocities (see Figure A3-10). The higher spread in stopping Forces compared with Starting Forces will now be examined.

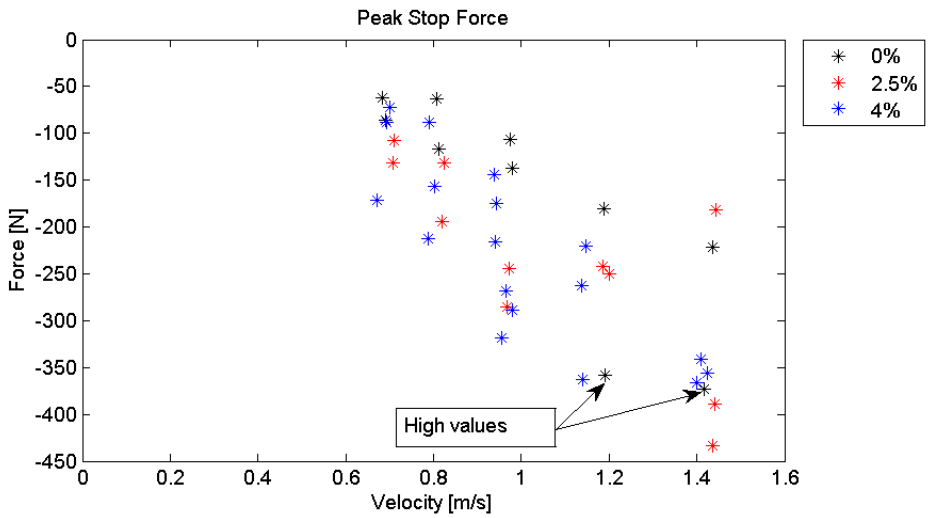

Figure A3-9: Individual peak Stopping Forces against velocity for each of the crossfall gradients, showing a general trend of increased magnitude of Stopping Force with crossfall gradient and velocity.

In general the peak force necessary to start the wheelchair moving was 3 to 5 times that of the average force needed to keep the wheelchair moving. This was apart from the fastest speed on the 0% condition where the peak start force was nearly 9 times the going force (see Table A3-7).

0% $\frac{F_{start}}{F_{going}}$	2.5% $\frac{F_{start}}{F_{going}}$	4% $\frac{F_{start}}{F_{going}}$	0% $\frac{F_{stop}}{F_{going}}$	2.5% $\frac{F_{stop}}{F_{going}}$	4% $\frac{F_{stop}}{F_{going}}$
3.48	3.24	3.27	3.08	4.38	2.97
4.16	3.77	3.51	3.25	5.71	4.28
4.53	3.64	4.63	5.04	9.56	6.84
5.08	3.41	4.28	12.82	7.44	7.41
9.07	5.16	5.2	17.71	10.91	9.61

Table A3-7: Ratios of peak start and stop forces to average going force for each velocity and each crossfall condition

The ratio of the peak force to stop the wheelchair to the average going force was much greater than the start ratio just described as the velocity of the wheelchair increased, reaching a peak of 17.71 on the flat condition. The reason for the higher ratios is probably due to the difference in methods used to start and stop the wheelchair. On the one hand the wheelchair was accelerated from rest by the power imparted by the scooter. However, the wheelchair was stopped by an attendant. The time taken to stop the wheelchair was less than that taken to start the wheelchair due in part to the difference in methods and in part to the fact we were aiming for as long as possible of constant velocity. Therefore the time available to stop the wheelchair was less than that to start it.

In conclusion the Starting Forces significantly increased with both crossfall gradient and velocity, and so the null hypotheses can be rejected. However, the test conditions were not robust enough to accurately measure the stopping Forces over similar conditions for all runs, which prevented a statistical analysis taking place and therefore the null hypotheses cannot be refuted for the stopping Forces. However, it can be noted that there was a general trend of increasing pulling force as velocity increased.

Having examined the effect of crossfall gradient on the four C_{RQD} (Starting Force, going Work, QSS Force and stopping Force) it can be concluded that there is an increase in force required to start the wheelchair (Starting Force) and to keep the wheelchair moving at a constant velocity (QSS Force). The result of the latter means there is an increase in going Work required once the wheelchair has started to move until it stopped. It would appear the stopping force was more dependent on the time and distance it was applied over than the crossfall.