

A micro-level approach to measuring the accessibility of footways for wheelchair users using the Capability Model

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There are a growing number of people with mobility impairments who use wheelchairs to get around the built environment. This number is likely to increase in the future due to an increasingly ageing population combined with advances in medical technology which help to overcome some of the barriers to access that have hitherto prevented people from leading as full a life as they would have liked. Footways form an integral part of the transport network and therefore it is essential they can be accessed by all people. Currently, however, there is no well-defined method to measure the accessibility of footways for wheelchair users. One aspect of a footway is the crossfall – the transverse gradient designed to facilitate surface water drainage – which adds to a wheelchair user’s difficulty when progressing along the footway. This paper first reviews previous research on measuring the effect of crossfalls on wheelchair accessibility, highlighting the need for a new approach. It then proposes the Capability Model as a starting point for this new approach. The model is updated and populated with an initial capability set chosen to measure footway accessibility across footways with three different crossfall gradients (0%, 2.5% and 4%). The focus is on the physical work provided by the user to the wheelchair in order to keep it travelling in a straight line. It is shown that in order to travel in a straight line when a footway is flat only a single principal capability is required: the ability to produce sufficient force over the required distance to overcome the inertia and rolling resistance and keep the wheelchair moving at the chosen velocity. When a positive crossfall gradient is introduced a second capability is required: the ability to apply different levels of force to the left and right sides of the wheelchair. It is concluded that it is possible to measure these two capabilities and these provide a good insight into the effect of crossfalls on footway accessibility for wheelchair users.

Keywords: wheelchair; footway; crossfall; cross-slope; accessibility; Capability Model

1. Introduction

Footways are a transport system in and of themselves. They also provide links to and between other transport modes. Therefore, being able to move along them is an essential skill to ensure a person’s mobility. An essential component of footway design is the crossfall – the transverse gradient used to aid drainage. This affects every user of the footway, and it is a particular issue for those whose pedestrian movement involves the use of wheels – e.g. pushing buggies – but it is a major problem for people in

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wheelchairs. There are at least 1.2 million wheelchair users in England (National Health Services Modernisation Agency [NHSMA] 2004), 45% propel themselves in a manual wheelchair and 34% are propelled in their manual wheelchair by an attendant (Sapey, Stewart, and Donaldson 2004). This paper considers only the self-propelling wheelchair users – attendant-propelled wheelchair users have different factors to deal with in comparison to self-propelled wheelchair users and these are being considered in separate articles which are in preparation.

Measuring the accessibility of footways for self-propelling manual wheelchair users depends on the interaction between the capabilities of the user, the type of wheelchair they are using, the weight of the wheelchair system and the design of the footway. It is therefore necessary to develop a robust method for assessing each of these independently and in combination.

A crossfall will naturally turn any wheeled device downslope (Brubaker, McLaurin, and McClay 1986; Department for Transport 2004) because of the influence of gravity. The turning moment will depend on the distance of the centre of mass from the axis of the rear wheels, the mass of the system and the friction between the wheels and the footway surface. It is facilitated by the presence of free-turning front wheels (casters) (Brubaker, McLaurin, and McClay 1986; Tomlinson 2000). This turning moment must be overcome if the wheelchair is to travel in a straight line.

Current UK footway guidelines state that the crossfall gradient should not exceed 2.5% because a gradient steeper than this could hinder the accessibility of wheelchair users and others with mobility impairments (Department for Transport 2005, 2007). The UK Highways Agency also acknowledges the difficulty crossfalls pose to those with mobility impairments, stating that crossfalls greater than 3% can be uncomfortable to walk on (Department for Transport 2004). They recommend a crossfall of between 2% and 3.3% with an absolute minimum of 1.5% and a maximum of 7%. The effect this has on wheelchair users as recorded in the published literature is contradictory. Some believe it poses little problem and the guideline could be relaxed (Kockelman, Zhao, & Blanchard-Zimmerman, 2001, 2002; Longmuir et al. 2003), while others have shown that it increases the physiological energy cost (Brubaker, McLaurin, and McClay 1986) and results in users needing to apply higher forces to the wheels (hand-rim forces) in order to obtain and control motion (Richter et al. 2007). These contradictory findings could be explained by the different approaches to studying the problem.

One approach emerges from the process of rehabilitation of wheelchair users following injury. The rehabilitation world is interested in the person and in restoring and maximising function while reducing the risk of associated injury. The primary mechanism for generating motion in a self-propelled wheelchair is placing the hand on the hand-rim and applying a forward or backward force which then rotates the wheel and thus provides linear motion as the wheel interacts with the floor surface. However, handrim forces applied to wheelchairs have been shown to be linked to upper-limb injuries (Boninger et al. 2004) and shoulder and wrist injuries are a particular problem for self-propelled wheelchair users (Bayley, Cochran, and Sledge 1987; Alm, Saraste, and Norrbrink 2008; Yang et al. 2009). Therefore, rehabilitation research, such as the work conducted by Richter et al. (2007), has concentrated on measuring the handrim forces (and other biomechanical properties) generated by a person propelling a wheelchair on a crossfall. Richter et al. subsequently recommend wheelchair users avoid crossfalls where possible to ensure they do not increase their risk of upper-limb injury. They further recommend future research to investigate the forces applied to the upslope side of the

wheelchair as well as the downslope side. However, virtually every outdoor journey involves the use of a footway and, as noted above, every footway has a crossfall, so avoiding crossfalls is not practical. Also, the research of Richter et al. does not answer the question of how a crossfall affects the accessibility of a footway though it does suggest that there may be a problem for wheelchair users when travelling along a footway with a crossfall.

In contrast, Kockelman et al. (2001) were interested in answering the question of accessibility. They had previously highlighted the lack of evidence for the then current guidelines for crossfall (referred to as 'cross-slope') gradients in the USA (Kockelman et al. 2000). These standards – set out in the Americans with Disabilities Act (ADA) Accessibility Guidelines (US Access Board 1994) – stated that a maximum crossfall gradient of 2% must be maintained on footways to ensure their accessibility for wheelchair users. Kockelman et al. carried out two sets of experiments, each using the same methodology. In the first study, 19 subjects with varying mobility impairments were used (Kockelman et al. 2001); in the second, the original 19 were combined with an additional 50 people, again with mixed mobility issues. In both cases, each participant's heart rate and opinion of how difficult a footway was to negotiate were used to populate a random-effects model and an ordered-probit model, respectively (Kockelman et al. 2001; Kockelman et al. 2002). In the first of these studies, 20 people participated in the study, 8 of whom used a powered wheelchair or electric scooter and 5 of whom used a manual wheelchair (Kockelman et al. 2001). This study concludes that a 4% guideline for the maximum crossfall gradient should be applied in order to maximise the accessibility of footways. It further recommends that when this is not possible to achieve this guideline the crossfall gradient can increase to 10% provided the longitudinal slope does not exceed 5% (Kockelman et al. 2001). It should be noted that this figure was arrived at assuming that 20% of people with a disability will find the crossfall 'difficult to cross'. Due to the practical difficulties of carrying out experiments on actual footways, it was difficult for Kockelman to find stretches of footway with a consistent crossfall gradient and without a longitudinal gradient. Therefore, there were different amounts of longitudinal slope on each of the test crossfall gradients. Vredenburg et al. (2009) also tested combinations of longitudinal slopes and crossfall gradients on 6 m (20 ft) plywood ramps (Vredenburg et al. 2009). They measured each wheelchair user's perceived level of exertion and perception of actual gradient. They recommend a crossfall gradient of 5% when the longitudinal gradient is 2% or less.

Neither Kockelman et al. nor Vredenburg et al. tested the difference of a crossfall gradient independently of a longitudinal slope. Therefore, the question still exists: do crossfalls form a barrier which inhibits the accessibility of footways for wheelchair users? To answer this question it would be necessary to look at the interactions between the three elements: the user, the wheelchair and the environment. One way of doing this is to develop a model which examines the contributions made by the wheelchair, the user and the environment independently as well as in their many possible interactions. One model that illustrates this is the Capability Model. This will now be discussed in detail.

2. Measuring accessibility with the Capability Model

2.1. Development of the Capability Model

The Capability Model was developed as a way of measuring the accessibility of the built environment. It is presented in detail in Cepolina and Tyler (2004) and Tyler (2006), and

is summarised here for convenience. The Capability Model has at its core two sets of measurements: provided and required capabilities. The former are all the things a person is able to do; the latter are all the things one would need to be able to do in order to complete an activity. Each activity is divided into tasks; each with its own set of capabilities. As the number or level of required capabilities increase, the task is deemed to be more difficult, as it will demand a greater level of provided capabilities in order for it to be achieved. This interaction is shown in Figure 1 which is adapted from Cepolina and Tyler (2004). This figure shows that when the provided capabilities are less than the required capabilities the person fails to achieve the task because the task presents more difficulties than the person can cope with. Thus, capabilities have multiple levels, and these can be mapped to the World Health Organisation’s International Classification of Functioning (ICF).

The ICF model is a departure from both the more traditional ‘medical model’ of disability, which focused on ‘fixing the medical conditions used to define disability’ and the ‘social model’ of disability, which took away all responsibility from the medical conditions and instead ‘blamed’ society’s inability to deal with the environment for making people disabled. The ICF model instead tries to look at what people can actually achieve: their functionings. For this reason, it has been described as the model which is the ‘closest to the definition of disability found under the capability approach [of Sen]’ (Mitra 2006). In the ICF model, there are seven domains of functioning. These are Health Condition, Body Function and Structures, Activities, Participation, Personal Factors and Environmental Factors.

The Capability Model is concerned predominantly with the Body Functions and Structures, Activities and the Environment domains of the ICF model. However, the Capability Model wishes to look at the interactions between these three domains, as opposed to categorising them.

A provided capability can be a sensory, mental or physical attribute of the person. Even a simple task such as climbing a step requires a plethora of capabilities and therefore initial studies using the Capability Model have used time as a proxy measure which represents the result of the interaction of all capabilities with the task in question. As an example time has been used as a measurement of how well a person is able to navigate through a maze as a means for testing the effects of a gene therapy intervention (Jacobson et al. 2012). This is described in terms of the Capability Model by Tyler (2011).

The use of time as a proxy allows researchers to test people with varying levels of mobility using a common factor; it does not, for example, exclude those with visual

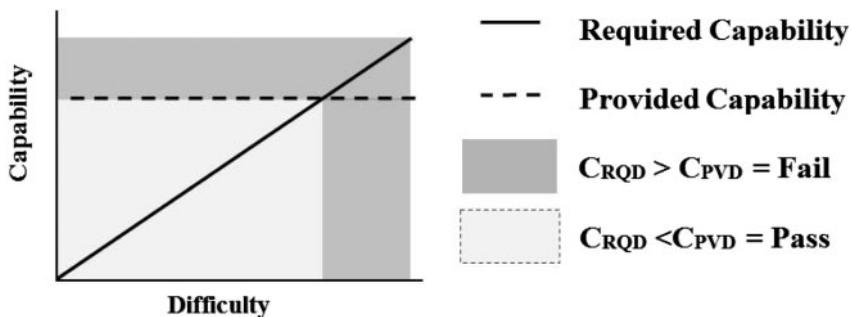


Figure 1. The Capability Model adapted from Cepolina and Tyler (2004).

impairments or wheelchair users. This gives it a great deal of power. There are, however, drawbacks to such an aggregate term. It does not reveal which capabilities are being used, when the choice of capabilities changes or when a person may be nearly failing – this could be marked in some circumstances by an increase in the time taken to complete the task, and in others by a decrease. There is also no way of identifying when completing a task may be causing injury. This is a particular problem for wheelchair users who are frequently at risk of upper-limb overuse injuries (Nichols, Norman, and Ennis 1979; Gellman et al. 1988; Sie et al. 1992; Veeger, Van der Woude, and Rozendal 1992; Oesterling et al. 1995; Curtis et al. 1999; Boninger et al. 2004) which can in extreme cases leave them unable to self-propel their wheelchair.

Measuring required capabilities requires a conversion from, for example, a physical or sensory measurement (e.g. height of a step, lighting level) to what this means in terms of capabilities – a given step height requires, amongst other things, the capability to raise the foot by at least that amount in order to overcome the barrier it presents.

The Capability Model can be adjusted to include people who use assistive technologies. An Assistive Technology is defined here as ‘any product or service designed to enable independence for disabled and older people’. [This definition was developed in the UK in 2001 to replace the term ‘disability equipment’ (see: <http://www.fastuk.org/about/definitionofat.php>).] The addition of the assistive technology introduces another two sets of interactions: between the user and the technology and between the technology and the environment. How this can be included in the Capability Model can be seen in Figure 2.

Wheeled mobility fundamentally changes the ways in which people can complete a task, compared to the methods required for walking. As such they (wheelchair users, scooter users and small children in pushchairs) are a distinct subset of the pedestrian population. Self-propelled wheelchair users are a special case of this subset because, unlike the other groups, the application of force to move or control the wheelchair is intermittent as it is necessary to grip and then release the handrim to provide the linear

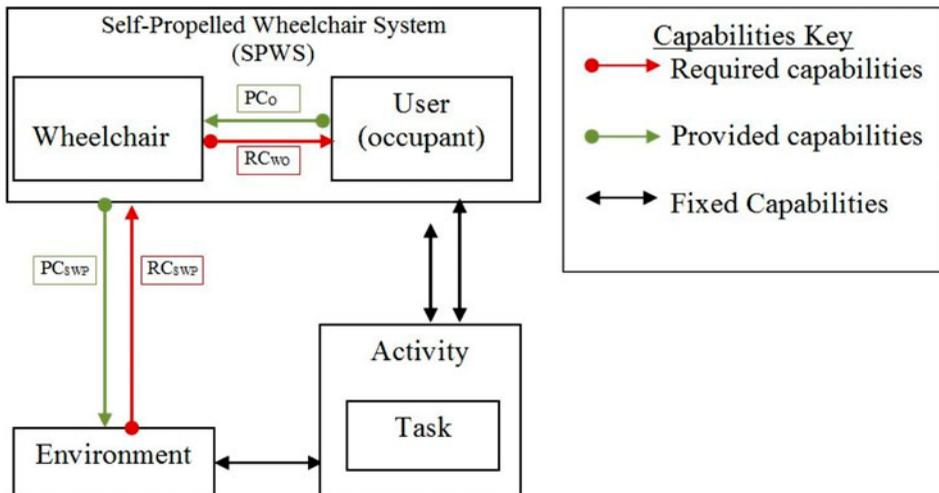


Figure 2. The Capability Model showing the interaction between the user and wheelchair (which form the SPWS), the environment and the activity.

movement. The fact that users must release the handrim means that the wheelchair is, periodically, uncontrolled and thus able to roll down slopes unless some control is applied. The Capability Model will now be adapted to investigate the effect of a crossfall on the accessibility of self-propelled wheelchair users.

2.2. Measuring wheelchair accessibility using the Capability Model

Measuring wheelchair accessibility means measuring the impacts of the interactions between the required and provided capabilities. These interactions will now be investigated.

The provided capabilities of a person are what they are able to do. There is a subtle but important distinction between the characteristics of a person and their capabilities. Their characteristics such as their age or gender simply describe them. What the capabilities model is interested in is what people are able to do. Therefore, characteristics are important in that they may help predict capability, but are not variables in the proposed capabilities model. In much the same way the required capabilities are taken to be the combination of the effect of these characteristics on people. Therefore, the Capability Model is actually about mapping the interactions between the person and the wheelchair and the environment for any given task. These interactions are detailed in [Figure 2](#), with required capabilities shown with red arrows and provided capabilities with green arrows.

Wheelchair–user interactions are often the domain of rehabilitation clinicians; they attempt to change either the wheelchair or the wheelchair set-up to reduce the required capabilities of the wheelchair. At the same time training is given to people to increase their provided capabilities. The outcome of these interactions results in the self-propelled wheelchair system (SPWS) having a provided capability (which is a function both of the user and the wheelchair). It is this system which will need to be able to access the built environment. Therefore, for civil engineers, it is the provided and required capabilities which exist between the SPWS and the environment which are of most interest; not least because, as civil engineers, they have some remit to change the environment. The provided and required capabilities are also dependent on the activity chosen by the person and the environment in which they must (or choose to) carry out the activity. For the purposes of this paper, these capabilities have been fixed; the task is to travel in a straight line along a footway both with and without a crossfall.

3. The effect of crossfall gradients on wheelchair users' accessibility

Self-propulsion of wheelchairs is, in general, achieved by the user imparting a periodic force bilaterally to each wheel by means of the handrim. How a user imparts this force to the handrim can vary based on their abilities and also on the design and set-up of the wheelchair. Regardless of how good (or bad) a person's technique the amount of work done in moving a wheelchair system of a given mass at a certain velocity will be the same. This applies only to the force applied in the tangential forward direction of the handrim as this tangential force component is the only component capable of moving the wheelchair forwards (see [Figure 3](#)).

3.1. Experimental methods

3.1.1. Hypothesis

The null hypothesis being tested in this paper is that there is no difference in the number of provided capabilities used by people as they self-propel wheelchairs on a footway with

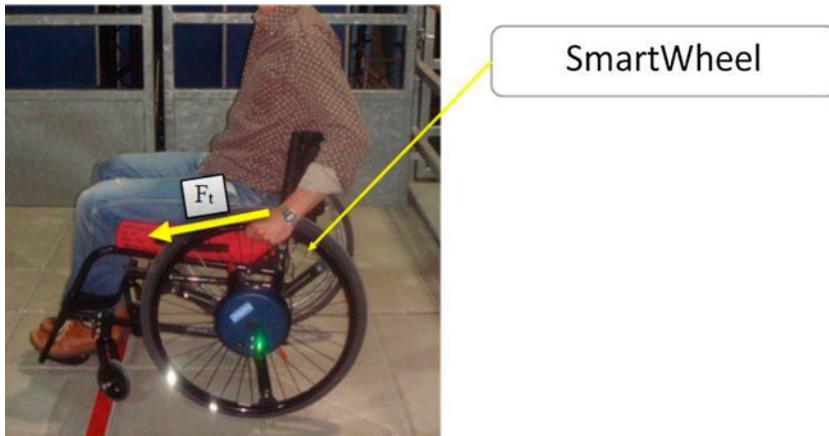


Figure 3. The Quickie™ wheelchair and SW used in the study showing the tangential force component (F_t).

a 0% crossfall compared with those with a positive crossfall gradient of 2.5% and 4%. The alternative hypothesis is that there is an additional capability provided by people as they push a wheelchair over a footway with a 2.5% and 4% crossfall compared with one which has a 0% crossfall.

3.1.2. Facility and equipment

The Pedestrian Accessibility and Movement Environment Laboratory (PAMELA) facility was set-up so that it contained three lanes of footway (10.2 m long and 2.4 m wide). Each footway had a different crossfall gradient: 0%, 2.5% and 4%. The surface was constructed with standard concrete pavers. The set-up is shown in Figure 4, which also shows the start and finish line as well as the dashed line placed on the surface to indicate the ideal travel path along each lane.

The SmartWheel (SW), developed by Three Rivers Holdings, is a commercially available wheel that can be fitted to a wheelchair in place of a standard wheel. It is capable of measuring three-dimensional forces and moments applied to its handrim, as well as the velocity of the wheelchair (see Figure 3). Details of the SW technical parameters can be found in Cooper (1997). The parameters of interest in the present study were the tangential force and the distance, which are calculated by the SW. The amount of work done in moving the wheelchair was calculated by integrating the tangential force with respect to distance travelled over each contact period.

The wheelchair used in this study was a Quickie GPV. This type of wheelchair is light-weight and is frequently provided by the National Health Service in the UK for 'active' users (people who use their wheelchair daily to access the outside environment). The wheelchair (shown in Figure 3) had a 63.5 cm standard wheel on one side and the SW on the opposing side. Both sides had a solid tyre to eliminate errors from differences in air pressure. The SW (shown in Figure 3) was always put on the side of the wheelchair coinciding with the non-dominant hand of the participant. The wheelbase of the wheelchair was 41 cm and the rear wheels were set-up with a camber of 2°. The casters of the wheelchair were solid and had a diameter of 12.7 cm.

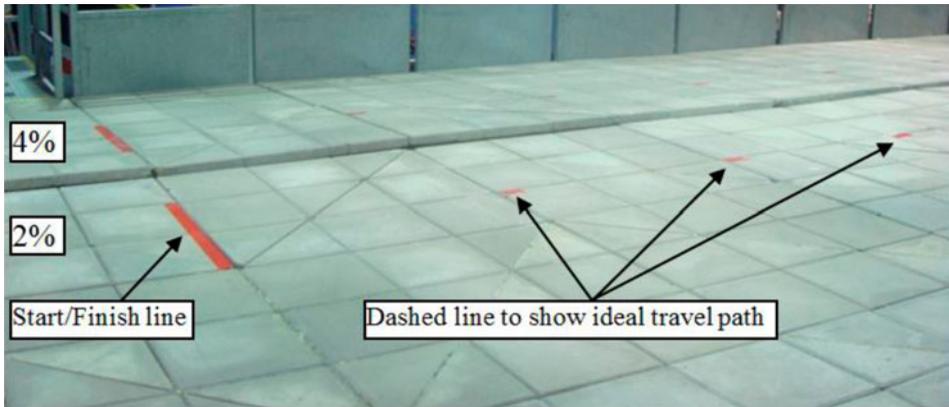


Figure 4. PAMELA set-up showing start/finish line for the experiments along with the dashed line participants were asked to follow.

3.1.3. Participants and protocol

Twelve able-bodied people were recruited for the study along with two regular wheelchair users. Participants were asked to sit in the wheelchair and spent some time on the PAMELA platform practising propelling the wheelchair until they felt comfortable using it. They were then, assigned a lane at random. Each person started behind the red start line and was instructed to follow the red dashed line and to stop after the stop line at the end of the lane. As the wheel could only measure the forces on one side of the wheelchair participants went up and then down each lane so that ‘upslope’ and ‘downslope’ work could be measured. This was repeated three times for each lane condition. All participants were asked to attempt all lane conditions in this way.

3.1.4. Data and statistical analysis

The data were trimmed to exclude the first and last push which were responsible for starting and stopping the wheelchair respectively (see Figure 5). Contacts with the handrim were then identified by finding local maxima peaks (the black stars in Figure 5) and local minima peaks (the grey stars in Figure 5) of tangential force. An inside-out search was then used to find where the tangential force dropped below (in the case of a positive contact) or rose above (in the case of a negative contact) 0 N. These times were recorded and the work done was calculated for each contact. The first and last contacts were excluded as they represented starting and stopping the wheelchair respectively. The work done by each remaining contacts was summed to give a value of the total work done in moving the wheelchair forwards.

The two capabilities (work done and work difference between left and right sides) were calculated by summing successive runs (an upslope and a downslope) on the same surface to get the ‘sum of work’ (C_{wk_sum}) and subtracting the absolute value of the downslope from the absolute value of the upslope to get the ‘difference of work’ (C_{wk_diff}).

$$C_{wk_sum} = W_{dn} + W_{up}$$

$$C_{wk_diff} = |W_{dn}| - |W_{up}|$$

All analyses were carried out using custom scripts in Matlab Version 7.11.0.

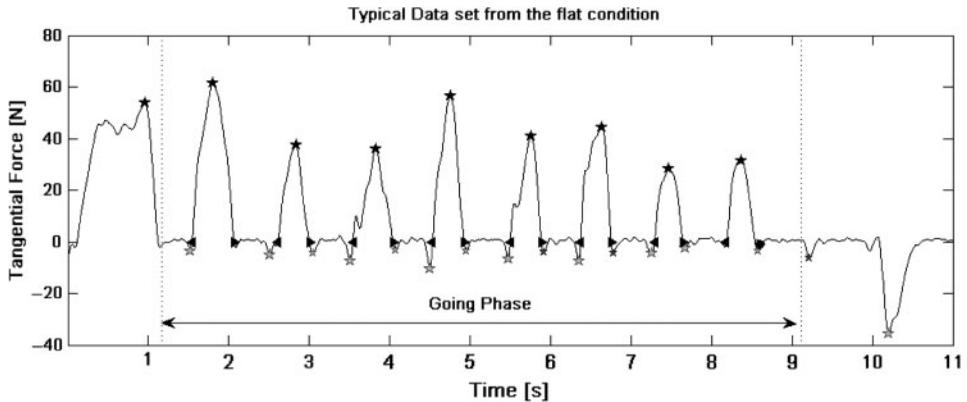


Figure 5. Figure showing a typical plot of tangential force against time. The start and end of each contact with the handrim is shown with a black triangle. The local maxima are shown with a black star and the local minima with a grey star.

3.1.5. Statistical analyses

The values for C_{wk_sum} and C_{wk_diff} were checked for normality using the Shapiro–Wilk test and then analysed using multiple linear regression analyses with occupant mass and crossfall gradient as regressors. A Bonferroni adjustment was applied to the significance level, which resulted in a significance level of $p = 0.017$. All statistical analyses were carried out using PASW Statistics 18, Release Version 18.0.0.

3.2. Results

3.2.1. Participant details

The participants were mainly male (12) with only two females. This gender imbalance was not designed into the experiment, but was a consequence of the difficulty found in recruiting females to take part in the study. Two participants were regular wheelchair users. These were originally recruited so that a comparison could be made with the able-bodied participants. However, as their data proved to be unremarkable both able-bodied and wheelchair user results were combined. The average weight of the participants was 69.08 kg with a standard deviation of 14.86 kg. The average age was 34.6 years (± 9.9 years). Eleven out of the 14 participants were right handed.

3.2.2. Provided capabilities: sum and difference of work

The results of the linear regression are summarised in Table 1. The results show that the regression model for C_{wk_sum} was a very poor fit ($R^2 = 0.063$, $R^2_{adj} = 0.047$), and although the relationship is significant [$F(2,123) = 4.04$, $p = 0.023$], 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.

When a model has such a poor level of fit this means that it is no better at describing the data than simply using the mean of the data. This is confirmed when one looks at the mean values for each condition as they 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.

Table 1 also shows the results of the multiple linear regression model for the effect of crossfall gradient on the difference of work (C_{wk_diff}). The model has a good degree of fit

Table 1. Regression model summary for the provided going work (C_{wk_sum}) and the difference of work (C_{wk_diff}).

Dependant variables	Model		Coefficients					
	R^2 (R^2_{adj})	p	Construction	p	Crossfall (%)	p	Occupant mass (kg)	p
C_{wk_sum}	0.063 (0.047)	0.02	71.79	<0.0001	-2.66	0.093	0.42	0.023
C_{wk_diff}	0.865 (0.863)	<0.0001	-51.12	<0.0001	-20.17	<0.0001	0.68	<0.0001

($R^2 = 0.865$, $R^2_{adj} = 0.863$) and was significant at explaining the variation in the data [$F(2,123) = 388.914$, $p < 0.0001$]. When the individual variables are examined 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 20.17 Nm ($p < 0.0001$). Occupant mass was also positively correlated to C_{wk_diff} ($p < 0.0001$) but the actual influence was low, with an increase of 0.684 Nm for every kg increase in mass. The model can thus be formulated in the following equation:

$$C_{wk_diff} = -51.116 + 20.174(C) + 0.684(M),$$

where C is the crossfall gradient as a percentage and M is the mass of the occupant in kilograms. The C_{wk_diff} for each gradient relative to occupant mass is shown in Figure 6.

Figure 7 shows the difference of work against the difference of work for each crossfall gradient. It can be seen there is a clear increase in the difference of work as crossfall gradient increases. The coping strategies employed by people when applying a difference of work while continuing to provide the work necessary to move the wheelchair ranged from those who simply reduced the magnitude of force on the upslope side, to those who had to apply frequent braking forces to the upslope side while increasing the magnitude of their pushes on the downslope side; with one pushing choosing to only push on the downslope side.

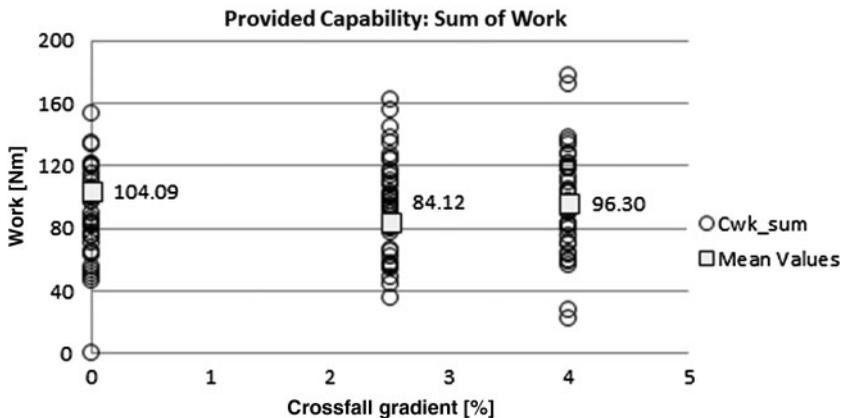


Figure 6. Sum of work (C_{wk_sum}) for the upslope and downslope sides of the wheelchair, showing no trend between amount of work and crossfall gradient. Mean values for each condition are displayed in red with the accompanying value.

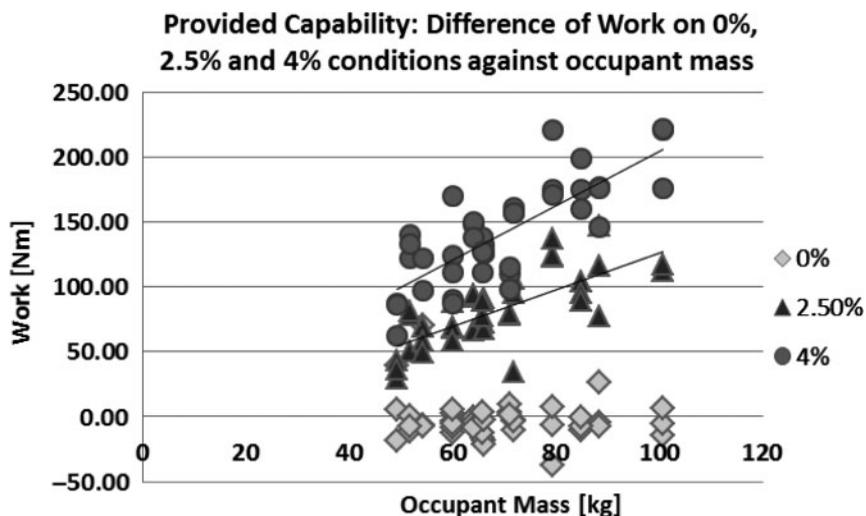


Figure 7. Difference of work (C_{wk_diff}) for 0%, 2.5% and 4% crossfalls plotted against occupant mass.

Discussion

This study set out to see if the Capability Model could be used to measure the effect of crossfall gradients for wheelchair users. Crossfall gradients were chosen due to the conflicting evidence regarding the impact they have on wheelchair users. This study is the first to quantify the difference of energy used (provided) by wheelchair users to travel a set distance over footways with different crossfall gradients. It is clear from our results that a second capability must be provided by a user if they are to overcome the downward turning moment.

The application of negative forces to prevent the wheelchair turning downslope was previously observed by Richter et al.; although they did not quantify the magnitude of the force, they recommended doing so in future studies (Richter et al.). Richter et al. did quantify a number of biomechanical measures for the downslope side of the wheelchair and showed that on average people travelled slower and needed to apply greater force to the downslope side of the wheelchair when travelling on a crossfall. However, they found that cadence (pushes per minute) remained unaffected. This contrasts to observations made during the current study with most participants, in which cadence was noted to change significantly as crossfall increased. This difference could be explained by one or both of the key differences between the studies: Richter et al. used an inclined treadmill and experienced wheelchair users whereas we used mostly inexperienced wheelchair users and the wheelchair was free to roll downslope. Despite this slight difference in findings, the main conclusions in both studies agree: crossfalls present a barrier to wheelchair users which necessitate the users to provide increased forces (either applied via increased push force to the downslope side of the wheelchair or braking force to the upslope side of the wheelchair).

The methods used in this paper are quite different from those of Kockelman et al. and Longmuir, which may account for the difference in results and conclusions. Neither Kockelman et al. nor Vredenburg et al. tested the difference of a crossfall gradient independently of a longitudinal slope due to the practical difficulties of finding isolated

examples of each this was something the present study was able to overcome due to the unique environment of PAMELA.

The recommendation for a 4% (10% in some cases) crossfall gradient by Kockelman et al. (2001) was arrived at by assuming 20% of people with a disability will find the crossfall 'difficult to cross'. We did not ask people how difficult they found the crossfall, which with hindsight would have made comparisons with the work done by Kockelman et al. easier. However, our research has shown a linear increase of the difference of work provided by people as crossfall gradient increases; and the necessity for all users to use at least one brake in lieu of a push on the downslope side. This linear relationship should remain if the crossfall gradient is tested at 10%, which would mean – according to our model's equation – an increase in the difference of work of approximately 200 Nm. To put this in context, Hurd et al. (2009) reported the amount of work done per push over smooth concrete to be approximately 3.5 Nm, over aggregate concrete they found 18.6 Nm was used and going up a 3 degree slope ($\approx 5.4\%$) this increased to 24.4 Nm.

In this paper, the Capabilities Model has been used to assess the accessibility of footways for wheelchair users. The advantage of this approach is that we have been able to identify what is provided by people to overcome the crossfall, this has been done independently of how each individual achieved this. In ignoring the 'how' we are able to define two broad capabilities required by the footway. However, we have ignored much of the biomechanical data. As an example of what has not been included at this stage, our definition of C_{wk_sum} is simply the work done to move the wheelchair; it ignores the additional forces applied to the wheelchair by the user which do not directly affect the forward motion of the wheelchair. The resultant amount of force provided by the user to the wheelchair is virtually always higher than the tangential force applied.

Conclusions

The Capabilities Model focuses on the individual's accessibility given a certain environment. This paper has expanded the model to allow for the inclusion of an assistive technology. The model allows for the capabilities provided by the individual to be identified, which in turn allowed us to make inferences about the capabilities required by the footway. It is therefore a distinct departure from the approach of Kockelman et al. whose recommendations were based on the population and from that of Richter et al. in which the detailed biomechanical analysis of forces and moments applied by the user to the wheelchair was the focus of attention.

In conclusion, when crossfall gradient increased the occupants had to change their pushing patterns on the upslope and downslope side of the wheelchairs. How they did this is beyond the scope of this paper but the fact is that they had to employ a second capability: that of applying a different force to the upslope and downslope sides of the wheelchair. This means that a crossfall makes propulsion along a footway more difficult. The results show that on average a person will have to create a difference of force resulting in a difference of work of approximately 50 Nm for a footway built to current UK construction standards with a 2.5% crossfall

The current study has expanded the Capabilities Model to allow it to be used to measure the effect of crossfall gradient on the accessibility of footways for wheelchair users. However, it has looked at a very limited set of capabilities and has not investigated how people provided the difference of work. These factors will be investigated in future studies to improve our understanding of the impact of crossfalls on wheelchair users.

Further areas of future research include applying the methods in this paper to: other surfaces, over longer distances, with experienced wheelchair users and with two SWs so that bilateral data can be collected simultaneously.

Crossfalls require the user to apply a difference of force to the upslope and downslope sides in order to keep the wheelchair moving in a straight line when moving on a footway with a positive crossfall gradient. This difference of work appears manageable within the limited number of individuals we tested when travelling on the 2.5% slope, however, all people had to begin to apply braking forces as the crossfall increased to 4%. It is, therefore, proposed that wheelchair users avoid steep crossfall gradients especially over long distances.

In order to ensure footways are accessible for wheelchair users it is proposed that footways are designed to within the recommended guideline of 2.5% wherever possible. This should remain the case when footways are widened in order to make streets more accessible, as making wider footways with steep crossfalls may inadvertently make the footway less, not more, accessible.

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