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Specification and simulation of erect passengers on public transport based on volunteer tests and real world data

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Executive summary

This deliverable presents the core activities of WP5 concerning standing passenger safety in public transport, with submitted manuscripts covering the different findings so far. The aim within WP5 is to demonstrate the application of Human Body Models (HBMs), specifically the open-source VIVA+ HBMs developed in WP2, with respect to standing passenger safety. The activity will provide a framework for future development of HBMs to enable injury prediction of standing passengers on public transport focusing on falls in normal transport modes, where e.g. emergency braking have been excluded.

Numerous factors increase the need for injury assessment tools to lower the risk of injuries to standing passengers on public transport, as these are susceptible to driver manuevers constituting of different magnitudes of acceleration and braking perturbations. Urbanisation, aging society, environmental sustainability and automated driving (AD) are all convincing motivators to ensure that public transport usage is safe, and standing passenger safety needs priority especially with upcoming development of AD systems in public transport vehicles. Fall injuries and impact scenarios with the internal structure of buses or trams can occur in free-standing postures such as boarding or alighting the vehicle, or when handles are out of reach. More knowledge concerning free-standing postural balance, when subjected to perturbations similar to those in public transport, is needed to guide safer operations and improve the safety of the passengers. Furthermore, this knowledge can be transferred to develop active HBMs, for both females and males, to enable simulation of standing passengers in public transport. The reference data for the development of active HBMs in WP2 was obtained from volunteer tests and will be demonstrated using the 50F VIVA+ HBM in Task 5.3. Conclusively, the knowledge from WP5 contributes to the development of a safe operation envelope for public transport with respect to standing passenger safety.

In this report, three manuscripts (two journal manuscripts and one conference short communication) covering the work based on the volunteer tests done at the University of Ljubljana, which have been submitted for publication will be described in brief (abstract, introduction, method, results, discussion, conclusion) with one chapter dedicated to each manuscript.

Another manuscript to be submitted to a journal is presented also. The first chapter will describe the volunteer tests and the main findings. The second chapter will describe balance strategies among the volunteers, based on video analysis from the volunteer tests, acting as a continuation of the work described in Chapter 1. Chapter 3 will describe further findings on the identified balance strategies from Chapter 2 and how kinematic data was derived, which will serve as input to the VIVA+ models in WP2. The fourth and final chapter describes a method to derive generic pulse shapes from recorded acceleration time series on buses, which can be used as a standardised framework in experiments and virtual testing to address standing passenger safety.

1 Human Response to Longitudinal Perturbations of Standing Passengers in Public Transport during Regular Operation

Simon Krašna, Arne Keller, Astrid Linder, Ary P. Silvano, Jia Cheng Xu, Robert Thomson, Corina Klug

This paper has been submitted to Frontiers Bioengineering and Biotechnology, section Biomechanics, for its special issue "Understanding Age and Sex-Related Differences in the Biomechanics of Road Traffic Associated Injuries Through Population Diversity Analyses". The final paper will differ from the manuscript presented here due to the review process.

This study investigates the response of standing passengers on public transport that experience perturbations during non-collision incidents. The objective of the study was to analyse the effects of perturbation characteristics on initial passenger responses and their ability to maintain balance. Sled tests were conducted on healthy volunteers of age 33.8±9.2 years (13 males, 11 females) standing on a moving platform facilitating measurements of initial muscle activity and stepping response of the volunteers. The volunteers were exposed to five different perturbation profiles representing typical braking and accelerating maneuvers of a public transport bus, in forward and rearward directions. The sequence of muscle activations in lower extremity muscles was consistent for the pulses applied. For the three acceleration pulses, combining two magnitudes for acceleration (1.5 & 3.0 m/s^2) and jerk (5.6 & 11.3 m/s3), the shortest muscle onset and stepping times to recover balance were observed with the high jerk value. The profile with the high acceleration magnitude and longer duration induced more recovery steps and a higher rate of safety harness deployment. A tendency for shorter response time was observed for the female volunteers. For the two braking pulses (1.0 & 2.5 m/s^2), only the low magnitude pulse allowed balance recovery without compensatory stepping.
The results obtained provide a reference data set for human body modelling, development of virtual test protocols, and operational limits for improving the safety of public transport vehicles and users.

1.1 Introduction

Passenger safety is necessary to ensure a sustainable transport system, to encourage even the most vulnerable persons to use public transport as a way of movement. Standing passengers are exposed to a risk of injury due to falling in non-collision incidents. Silvano and Ohlin (2019) identified that acceleration and braking results in different scenarios of passenger falls, e.g. acceleration and turning from the bus stop led to passengers falling after boarding while attempting to get seated. The main groups affected were older people (65+) and female passengers. In contrast, braking maneuvers affect males, females and different age groups similarly.

During perturbations, caused by e.g. typical driver manuevers and non-collision incidents, the human postural balance becomes unstable which increases the need for postural adjustments. During less severe perturbations, so-called *fixed-support strategies* are applied which are mainly ankle and hip joint motions to counteract perturbation forces acting upon the body. In this scenario, the body is within the same *base of support* (BoS), defined as the surface enclosed by the stance defined by the feet. During more severe perturbations, the center-of-mass (CoM) of the body is displaced outside the BoS, and to

maintain postural balance, the BoS is adjusted to contain the CoM within this area. This occurs through so-called *change-in-support strategies* which are mainly stepping strategies i.e. recovery/compensatory steps (stepping allows the BoS to move to contain the CoM, to counteract falling). Rogers et al. (2003) showed that shorter step initiation and completion time can be related to improved balance; i.e. to withstand perturbations, balance recovery stepping needs to be initiated quickly to reduce the risk of falling, the action of fast and effective compensatory stepping starts the recovery phase. However, multiple steps are a sign of balance instability and can result in large body displacements which increases the risk of impacting the interior of the vehicle (Robert et al., 2007; Siman-Tov et al., 2019; Zhou et al., 2020). To allow free-standing passengers to maintain balance, i.e. standing without hand support, Karekla and Tyler (2018) recommended acceleration and jerk thresholds of 2.0 m/s^2 and 0.9 m/s^3.

Human body models (HBMs) are mathematical models with high level of anatomical details developed to assess injury risks in different loading scenarios. To validate the HBM response to ensure biofidelity, necessary experimental data is needed. For injury assessment of free-standing public transport passengers, their responses in realistic conditions are needed. Furthermore, there is a need to provide guidance to drivers and developers of future autonomous vehicles to ensure standing passenger safety. Therefore, human postural balance during different perturbations on public transport needs to be further understood.

The aim of this study was to collect experimental data for the development of a standing passenger HBM to assess passenger responses, for both males and females, to different balance perturbations. The objectives were to (1) identify how different perturbations affect initial passenger responses, and (2) understand the consequences of the pulses with respect to initial responses between males and females.

1.2 Method

Note: A shorter, simplified description of the volunteer test setup and measurements is described.

Table 1.1 displays basic data on the 24 volunteers (13 males, 11 females) representing close to a 50th percentile anthropometry on average (body weight and height) that were exposed to five different perturbations in forward and rearward directions on a linear translational platform. To ensure volunteer safety, a full-body safety harness and a cushion on a location on the platform where a fall could have happened was used. The volunteers were perturbed from a stationary position (both feet on the moving platform, hip wide apart) with no information about the pulse characteristics. They were instructed to maintain a relaxed free-standing posture as they would as passengers on a bus. The corresponding perturbation profiles were severe enough to exceed published passenger comfort levels and also of magnitudes that are typical during regular travel at non-collision incidents. This would ensure that the volunteers were challenged to actively attempt to recover their balance. Furthermore, the pulses should be long enough to estimate whether the resulting body motions would put a real bus passenger at risk of colliding with the vehicle interior. Table 1.2 and Figure 1.1 illustrate the perturbation profiles and the sequence of their application, where *Br1* and *Br2* denote the braking pulses and combinations of *Acc*[1,2]-*J*[1,2] denote the different acceleration pulses, which differ in shape and can therefore evoke different muscle and kinematic response of the passenger, possibly influencing the risk of injury.

Table 1.1. Basic anthropometric and demographic data for the volunteers (mean±SD)

	Age	Mass	Height	Centre of mass height
	years	kg	cm	cm
11 females	31.6±7.2	64.7±9.9	165.5±6.4	91.2±3.9
13 males	35.5±10.6	86.2±11.8	179.2±5.4	99.2±3.6

Table 1.2. Perturbation profile characteristics

Profile name	Sequence	Magnitude	Rise time	Duration	Jerk	Displacement	Max. velocity
		m/s	s	s	m/s3	m	m/s
Br1	1	1.0	4.4	4.7	0.3	2.94	2.4
Acc1-J1	2	1.5	0.4	2.3	5.6	2.65	2.0
Acc1-J2	3	1.5	0.2	2.2	11.3	2.58	2.0
Acc2-J1	4	3.0	0.8	1.8	5.6	2.69	3.1
Br2	5	2.5	2.2	2.5	1.7	1.82	3.2

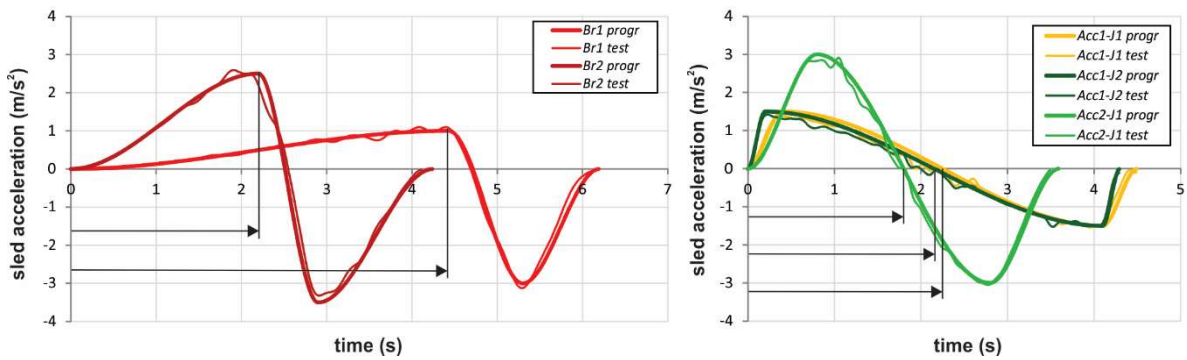

Figure 1.1. Perturbation pulses applied to the moving platform with standing volunteers; programmed pulses (thick lines), exemplary measured pulses (thin lines). Arrow lines indicate the parts of the pulses used in analysis of the volunteers' responses.

The study of the volunteers' response was limited to the initial rise time of the braking pulses and to the duration of the acceleration pulses, before the start of the sled deceleration to bring the platform to a stop. Nevertheless, the sled stopping segment raised safety concerns after pre-tests with *Br2*, where high magnitude of 3.5 m/s^2 was programmed due to the setup design limitations (Figure 1.1). As the bus braking and acceleration pulses were simulated in the same sled direction, a forward-facing volunteer experienced the accelerations similar to a transit passenger facing the direction of travel,

whereas the braking pulses were experienced as if the passenger were facing backward in the vehicle, opposite the direction of travel. The opposite was true for the rearward facing passenger.

Two high-speed cameras captured the volunteer's motion in the sagittal and frontal body planes and muscle activity was measured with electromyography (EMG). Body segment motions were captured with a system of eight cameras Oqus 3+ (Qualisys, Gothenburg, Sweden) tracking 56 passive reflective markers attached on the volunteer's body.
The sequence of events during the balance recovery was identified from the high-speed video recordings, where up to four sequential steps were tracked. The timing of the first frame when the contact between the foot and the ground (the moving platform) was lost was identified as *contact-off* time, while the time of re-establishing the contact was identified as *contact-on* time. The difference between *contact-off* and *contact-on* for the same (swing) foot represented the *swing time*. If the volunteer's motion was restricted by the harness before the end of the pulse, the event was identified as *harness deployment*.

1.3 Results

Note: A simplified description of the main findings are provided. EMG results and detailed statistics have been left out for this report, but the most important conclusions from the data analysed are briefly described in the discussion.

Eleven volunteers finished the complete set of tests, seven volunteers repeated at least one of the tests, and the higher severity pulses were omitted for six volunteers due to safety considerations. In total, 223 out of 238 tests were included in the analysis. More than half (57%) of the volunteers needed at least one compensatory step to maintain balance for the 1.0 m/s^2 braking pulse (*Br1*) in forward direction and 96% needed at least one compensatory step for the corresponding rearward *Br1* pulse. The results regarding compensatory steps, up to four sequential steps as described in the previous section, are presented in Table 1.3.

Table 1.1. The percentage of sequential steps (1st–4th) during balance recovery and the percentage of harness deployment in forward and backward facing volunteers (Males/Females).

	Forward					Backward				
Profile name	1st %	2nd %	3rd %	4th %	Harness %	1st %	2nd %	3rd %	4th %	Harness %
Br1										
M+F	57	52	30	9	4	96	70	57	30	9
M	58	58	25	17	8	100	50	58	17	17
F	55	45	36	0	0	91	91	55	45	0
Br2										
M+F	100	82	71	24	0	100	100	83	39	11
M	100	73	64	18	0	100	100	91	36	9
F	100	100	83	33	0	100	100	71	43	14
Acc1-J1										
M+F	100	100	67	54	21	100	92	79	33	21
M	100	100	69	46	15	100	85	69	23	23
F	100	100	64	64	27	100	100	91	45	18

Acc1-J2										
M+F	100	83	70	48	17	100	78	43	17	22
M	100	77	62	23	15	100	75	25	8	25
F	100	90	80	80	20	100	82	64	27	18
Acc2-J1										
M+F	100	100	92	46	75	100	100	88	50	88
M	100	100	85	31	69	100	100	92	46	92
F	100	100	100	64	82	100	100	82	55	82

In brief, the main finding was that for the highest acceleration pulse (Acc2-J1), the safety harness was deployed for the majority of the volunteers, 75% in the forward direction and 88% in rearward direction. The contact-off time in *Acc1-J1* and *Acc2-J1* was longer than in *Acc1-J2* in both directions ($p<0.001$), however no significant difference was found between contact-off time in *Acc1-J1* and *Acc2-J1*. For the braking pulses, the analysis showed significant effect of pulse, with shorter contact-off time in *Br2* in both directions. The contact-off and swing times of the initial compensatory step are presented in Table 1.4.

Table 1.2. Average times for initiation (contact-off time) and duration (swing time) of the first step, where observed (Males/Females, mean±SD)

	1st step contact-off time		1st step swing time	
Profile name	Forward ms	Backward ms	Forward ms	Backward ms
Br1				
M+F	3358±434	3205±441	153±69	166±67
M	3351±447	3183±434	177±68	173±68
F	3366±460	3236±475	125±65	158±68
Br2				
(M+F)	1259±228	1239±234	168±58	177±57
M	1244±186	1220±263	177±63	181±62
F	1288±310	1269±197	152±46	171±54
Acc1-J1				
M+F	541±96	634±89	136±44	171±55
M	556±94	641±79	150±41	181±65
F	523±100	626±104	120±43	160±40
Acc1-J2				
M+F	408±29	505±71	147±57	172±44
M	418±30	528±65	150±65	183±49
F	393±21	476±69	143±47	160±36
Acc2-J1				
M+F	577±96	672±92	155±54	165±40
M	601±90	682±61	173±44	169±48
F	549±98	660±124	133±60	161±29

The initial response to the perturbations followed the same pattern for the male and the female volunteers, but the results implied faster response for the females. Furthermore, the muscle onset latency, contact-off time, and swing time were found to be correlated to the body mass distribution.

Hence, the lower body mass and the lower CoM of the female volunteers (Table 1.2) could contribute to the faster response.

1.4 Discussion

A strong individual variability was observed during the tests: while some of the participants showed good ability to counteract the perturbation pulses used, others could not be exposed to the more severe perturbations for safety reasons, which also resulted in missing observations that could not be included in the analysis. The results of initial responses to different perturbations provides necessary information for the development of active HBMs for standing passengers on public transport.

In both directions of travel, the time until the participants initiated the first recovery step was longer for the braking pulses *Br1* and *Br2* than for the acceleration pulses (Table 1.2). The participants could maintain balance without recovery stepping in about half of trials with the low severity braking pulse *Br1*, while at least one recovery step was needed in the acceleration pulses. This is in accordance with observations in a previous study (Schubert et al., 2017), where volunteers (age 68.1±5.2) had to make recovery stepping when standing freely in a bus and subjected to accelerating and decelerating manoeuvres comparable to *Acc1-J1* and *Acc1-J2* pulses, while the magnitude of the deceleration phase was between *Br1* and *Br2* pulses. These authors found characteristic patterns of muscle activity similar to the observations in our study and observed a correlation between jerk and fast compensatory steps. In addition, the current study presents a more detailed analysis of the timing of muscle activity and stepping, which is needed for HBM validation.

In general, the pattern of muscle activation was similar for all the pulses, despite being considerably longer for the braking than for the acceleration pulses. The muscles activated were the opposite between the forward and rearward pulses, with anterior leg muscles preceeding the activation of posterior leg muscles during the forward pulses, and vice versa for the rearward pulses. The ankle strategy was pronounced before the initial compensatory step, in both directions. The higher jerk (*Acc1-J2*) magnitude was the most important factor in evoking rapid reflex responses, which was evident when comparing *Acc1-J1* to *Acc2-J1* that had the same jerk magnitude, but the jerk appeared at 0.1 s for *Acc1-J1* compared to 0.2 s for *Acc2-J1*. Hence, a higher jerk magnitude seems to evoke faster recovery stepping than a higher acceleration magnitude, within the range of magnitudes in this study.

Rearward stepping in response to a forward motion of the platform was consistently faster than forward stepping, which can most likely be attributed to the asymmetry of the human body in the sagittal plane resulting in different motion patterns for forward and rearward displacements (Runge et al., 1999). The percentage of participants that needed at least one recovery step was higher in rearward trials (Table 1.1), which is consistent with previous estimations of the single-step threshold being higher for forward direction, about 1.0 m/s^2, and lower for rearward direction, about 0.7 m/s^2 (de Kam et al., 2017).

The rate of harness deployment indicating excessive whole-body displacement was greater when travelling rearwards than forwards and particularly high in the *Acc2-J1* pulse (88%). An acceleration magnitude of 3.0 m/s^2 caused almost 90% of the rearward facing participants to fall into the harness, compared to 21% in *Acc1-J1* (which had the same jerk level, but half the acceleration magnitude). Although these findings cannot be directly compared to previous studies due to different setup and design of the safety system in the tests, the sharp rise of the harness deployment between 1.5 and 3.0 m/s^2 acceleration magnitude supports the comfort threshold levels of 1.0–1.8 m/s^2 as recommended in public transport (De Graaf and Van Weperen, 1997; Szturm et al., 1998). The percentage of harness deployment was low in the trials with the *Br2* pulse, even though it was identified as the most challenging to participants during pre-tests. Therefore, for safety reasons, the *Br2* pulse was not used for some of the volunteers during the tests. Moreover, the safety harness deployed during the sled stopping in the *Br2* trials, while the volunteers' response was observed during the rise segment (2.2 s;

Table 1.2, Figure 1.1) of the pulse only. In 24% of the forward trials and 44% of the backward trials, the harness was deployed after the rise time of *Br2* pulse ended (2.2 s, **Error! Reference source not found.**), before the stopping of the sled had to be initiated due to the operational limits of the setup. Compared to a free-standing posture, use of handrails and vertical bars substantially increases the possibility of standing passengers maintaining balance (Robert et al., 2007a; Sarraf et al., 2014; Schubert et al., 2017). However, if public transport must accommodate free-standing passengers, the vehicle acceleration and braking actions should be such that they minimise the risk of these passengers losing balance. Therefore, a starting point would be to limit peak accelerations to less than 1.5 m/s^2 and deceleration to less than 1 m/s^2. The jerk levels (although higher than recommended for comfort travel, above 5 m/s^3) did not cause major balancing issues and allowed the volunteers to recover balance effectively, with the room for compensatory stepping.

1.5 Conclusion

The response of standing passengers on public transport was captured by testing several perturbation profiles based on real-world recorded data; the shape, magnitude, and duration of the pulses significantly affected the initial response of free-standing passengers. Bus acceleration might induce a higher risk of passenger falling than braking due to higher jerk magnitudes as observed in this study. A combination of jerk and acceleration magnitudes needs to be considered when analysing balance responses in virtual testing with generic perturbations. Gender differences in initial step response were found and imply that gender-specific human body modelling can improve the HBM biofidelity and injury risk assessment for standing public transport passengers by means of virtual testing.

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2 Identifying and characterising types of balance strategies among females and males to prevent injuries in free-standing public transport passengers

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This paper has been submitted to Frontiers Bioengineering and Biotechnology, section Biomechanics, for its special issue "Understanding Age and Sex-Related Differences in the Biomechanics of Road Traffic Associated Injuries Through Population Diversity Analyses". The final paper will differ from the manuscript presented here due to the review process.

Free-standing passengers on public transport are subjected to perturbations during non-collision incidents caused by driver maneuvers, increasing the risk of injury. In the literature, the step strategy is described as a recovery strategy, when the center of mass is perturbed and the base of support is displaced, as a consequence of a severe perturbation. A differentiation of balance recovery strategies aimed at lowering the injury risk of free-standing public transport passengers have been characterised and evaluated within the current study. High-speed video recordings of 24 healthy volunteers (11 females and 13 males) subjected to five different perturbation profiles in forward- and rearward-facing translation, utilising acceleration, jerk, and braking maneuvers, have been qualitatively analysed. A methodology categorising a balancing reaction to an initial strategy and a recovery strategy, was used to identify and characterise the different balance recovery responses among the volunteers. A grading criterion was used to understand the effectiveness of a particular balance recovery strategy. The results show that the current definition of the step strategy is too primitive to describe the different identified recovery strategies. Stances were found to counteract continuous stepping more effectively than using a pure step strategy. The main prevalent and effective strategy was labeled the "fighting stance", where stepping was utilised effectively in this strategy by the top-ranking volunteers. High jerk perturbations seemed to induce faster and more successful balance recovery, mainly for those adopting the fighting stance, compared to the high acceleration and braking perturbation profiles. The findings might be useful in practice and for future studies, to lower the risk of injuries among free-standing passengers on public transport subjected to unexpected perturbations, and encourage in-depth qualitative analysis of balancing behaviors to provide more characterised details of human postural balance in the literature.

2.1 Introduction

Note: The introduction regarding public transport is similar to that in Chapter 1. Here, the shift of focus is a continuation of the study in Chapter 1 to understand what kind of balance strategies different volunteers might adopt.

The literature on human postural balance is extensive and often described using three fundamental balance strategies: 1) the ankle strategy, 2) the hip strategy, and 3) the step strategy (Nasher & McCollum, 1985; Winter, D. A. 1990). Another strategy found in the literature, though not extensively used, is the squat strategy, which incorporates both knee and hip flexion for stability (Cheng, 2016; Hemami & Pai, 2006). The ankle and hip strategies are fixed-support strategies, while the step strategy is a change-in-support (CIS) strategy induced during more severe perturbations as the center-of-mass

(CoM) and base-of-support (BoS) are displaced due to the momentum of the perturbation. The BoS is defined as the area under and between the feet. To maintain the full-body system in balance, the CoM of the body projecting on the floor must be within the BoS to maintain postural balance equilibrium.

For lower perturbations, the combination of ankle and hip adjustments is usually sufficient to maintain balance. For more severe perturbations, change-in-support strategies with single or multiple recovery steps are the most dominant strategies seen where no instructions during the recovery phase was given (Robert et al. 2012). However, as multiple stepping behavior can be executed in infinite ways, it would be interesting to understand the characteristics of different step strategies with respect to balance recovery. Is it possible to find characteristics of an effective balance recovery strategy, which could be implemented in practice to provide tips and tricks on increasing postural balance in free-standing scenarios when subjected to perturbations?

It is well-known that older people tend to execute shorter step lengths, have longer step initiation time, and need multiple steps compared to young people, all of which have been associated with less successful balance recovery (Do et al., 1982; Luchies et al., 1994; Thelen et al., 1997; Hsiao and Robinovitch, 1999; Wojcik et al., 1999). This makes biomechanically sense, as a larger step relocates the stepping foot ahead of the CoM and generates larger contact forces between the foot and the ground (King et al., 2005). There are numerous studies subjecting volunteers to unstable balance, e.g. through release experiments (to simulate trips and falls; the subject is released from different body angles and recovery stepping is evaluated) and translational perturbations in forward, rearward, and lateral direction. The recovery stepping is often analysed, usually differentiated by single-steppers, mixed-steppers (utilising single steps or multiple steps depending on which perturbation trial during the experiment), and multiple-steppers (Mille et al., 2005; Carty et al., 2012; Borelli et al., 2019). However, differentiation beyond that, with more details concerning identification and characterisation (description) of different stepping strategies that volunteers utilised during the trials, are rarely defined. Muscle responses, different step dimensions (time, length, frequency), and perturbation thresholds with respect to balance recovery are often defined, but it is interesting to understand how the execution of stepping strategies differ. Such investigation might provide insight on what characterises an effective recovery strategy, based on observed balancing responses.

Free-standing public transport safety needs to be accounted for to ensure successful balance recovery during non-collision incidents, when hand support might be out of reach or insufficient. Since young people have higher perturbation thresholds with respect to postural balance, subjecting these to more severe perturbations can provide guidelines on upper perturbation thresholds for free-standing passengers on public transport.

The aim of this study was to provide a first investigation on identification and characterisation of recovery strategies, beyond what is provided in the current literature, to provide insight on human postural balance during more severe perturbations. The idea was to differentiate the volunteers based on the different identified recovery strategies and qualitatively evaluate the effectiveness based on their balance recovery.

2.2 Method

The methodology used for the analysis of the balance strategies was based on visual analysis of video recordings of dynamic tests with volunteers, where the free-standing participants were exposed to translational acceleration perturbations (described in Chapter 1). The idea was to describe balancing reactions occurring during different magnitudes of acceleration, braking, and jerk, to provide insight on how balance recovery was executed and identify the perturbation magnitudes that caused problems with postural balance. The qualitative analysis comprised of three steps: (i) identification, aiming at

distinguishing different strategies used among the volunteers; (ii) characterisation, describing the execution and characteristics of the identified strategies, and (iii) evaluation, with the aim of systematically assessing the effectiveness of the identified and characterised strategies qualitatively.

To identify the different balance strategies, the perturbation trial was categorised into two phases:

1) *Initial phase:* indicated by the first balancing reactions occurring in the starting position when subjected to a perturbation. Hence, this is the first balance strategy executed by the volunteer. It is described by the fixed-support strategies, the ankle, hip, and squat strategies, since the BoS is stationary at this point (no stepping) and the CoM is displaced from its equilibrium at the start of the perturbation. If the perturbation is not severe, then balance can be maintained by a fixed support strategy.
2) *Recovery phase:* defined as the phase where recovery beyond fixed-support strategies (initial phase strategy) is induced through stepping, caused by more severe perturbations. The balance instability displaces the CoM and BoS beyond static equilibrium limits and a change-in-support strategy (step strategy) was utilised to keep the CoM within the translating BoS.

The hypothesis is that the step strategy defined in the literature is too primitive to characterise the differentiation of balance responses among the volunteers, i.e. the stepping responses can be differentiated (identified) and described (characterised) in more detail than dividing volunteers into single-, mixed-, and multiple-steppers. The aim of the evaluation of balance strategies was to understand the effectiveness of the identified balance recovery strategies. Here, a qualitative grading criterion was developed to rank the volunteer based on their balance recovery during the different pulse trials.

A grading scale of 2 (*effective*), 1 (*less effective*), or 0 (*ineffective*) was defined. If the volunteer managed to hold a stable position (stance) after stepping or managed to return to the starting position (fully controlled step strategy), then the strategy was considered "*effective*" due to successful postural control and effective stepping. The postural adjustments were deemed "*effective"* if it was evident that the volunteer had recovered balance, and displayed control throughout the compensatory stepping. In contrast, the strategy was deemed *"ineffective"* if the volunteer showed instability in the stance or postural adjustments, and exhibited an unstable balancing state, i.e., rigorous compensatory stepping representing difficulties in counteracting the perturbation, or harness deployment indicating that the recovery steps were not effective enough to recover balance. However, a strategy would be *"less effective"* if balance has been achieved yet showing some instability or constant utilisation of compensatory steps or adjustments throughout the perturbation with no clear equilibrium (equilibrium would be indicated by stationary positioning on the sled, i.e. no stepping). Displayed instability through compensatory stepping was considered to increase the risk of harness deployment and in contrast not ideal inside a public transport vehicle to avoid the risk of impacts due to more body displacement and less postural control.

It is well known that there are anthropometric differences between females and males, generally more evident in terms of height, musculature and fat mass, to name a few (Schneider et al., 1983; Al-Haboubi, 1997; Glenmark et al., 2004; Schorr et al., 2018). Physical capabilities, either through gender and anthropometrical differences or athletic background and experience, might affect the execution of a balance strategy. Thus, from the identification and evaluation of balance strategies, gender differences were examined to understand the effectiveness of utilised strategies and their execution. The success and failure rate of the volunteers due to the different pulse severities, were also estimated.

2.3 Results

From the qualitative video analysis, different strategies were identified to group the volunteers into specific balance strategy categories. Different execution of similar strategies was found among the volunteers but also between genders. The identified strategies were divided in an initial strategy and a recovery strategy. The overall initial and recovery strategies for a volunteer were based on their most frequently used strategy during the different perturbations. A success and failure rate presents how well the volunteers performed as a group, to denote the most challenging perturbation.

Overall, the main initial strategy was the ankle strategy. Knee flexion reactions (knee strategy) were in some cases found as part of the initial strategies. The hip strategy was also identified as an initial strategy, although less frequent than the ankle and knee strategies. For more severe perturbations, the step strategy was executed after a brief ankle strategy, displaying balance instability caused by the perturbation. The step strategy was the most prevalent strategy to recover balance as the BoS was displayed, with two main differences identified. The first one was the pure step strategy (pure steppers), denoting mainly continuous compensatory stepping and stretching the harness out, and the second one was identified as a counteraction to the stepping by utilising a stance to recover a stationary BoS (denoted as the *fighting stance*, Figure 2.1). Figure 2.2 displays the pure compensatory, continuous stepping. Figure 2.3 displays compensatory stepping with the aim of executing the *fighting stance*.

Figure 2.1. – Fighting stance in martial arts (left) and Volunteers 12 and 16 displaying their fighting stance (right). Left picture from (attached link): Two Male Mannequins Black White Fighting Stock Illustration 1877892802 (shutterstock.com) - Accessed 3 Feb 2021.

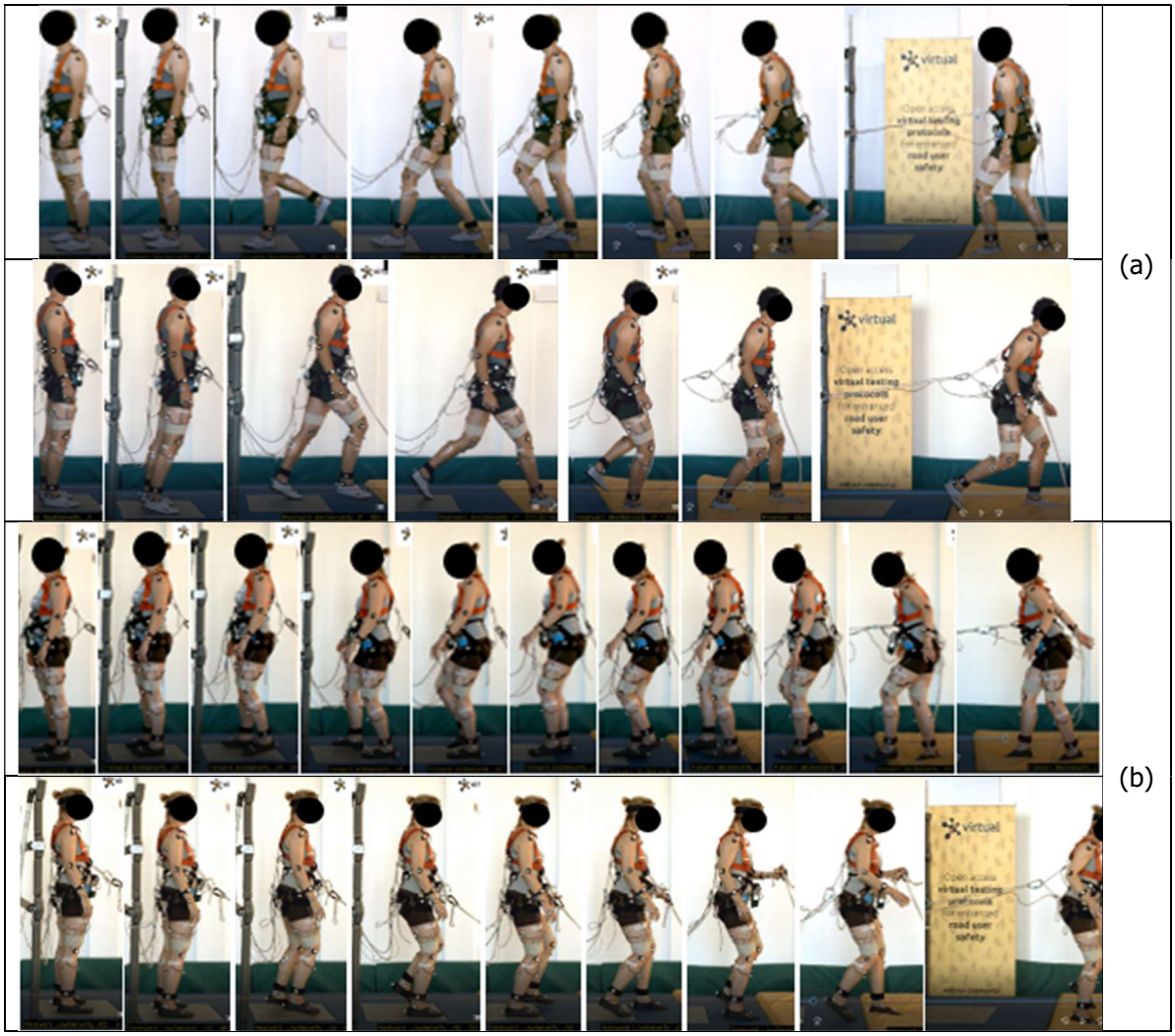

Figure 2.2. Typical forward and rearward step strategy, (a) male (Volunteer 22) and (b) female (Volunteer 10).

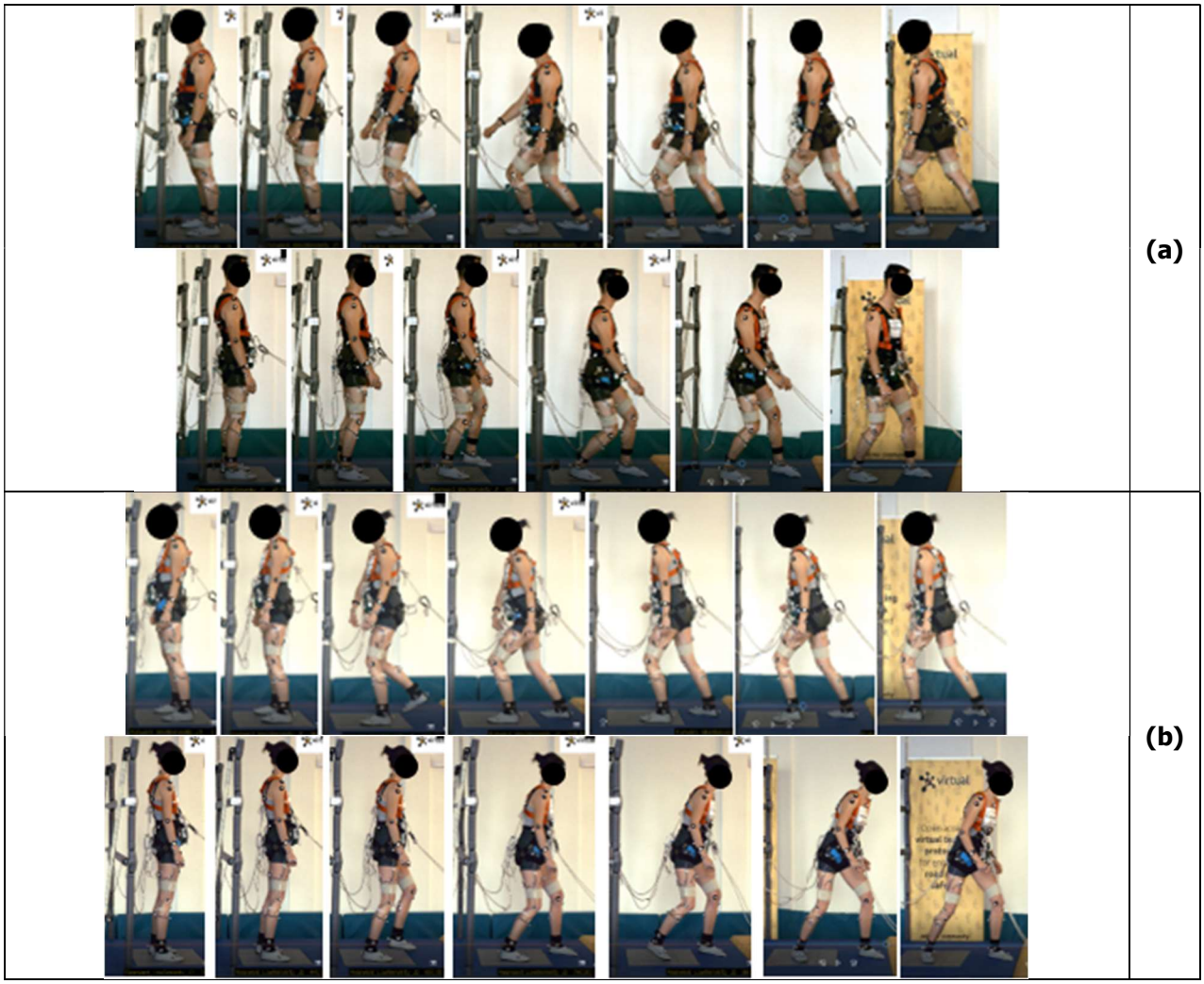

Figure 2.3. Typical forward and rearward fighting stance, (a) male (Volunteer 12) and (b) female (Volunteer 16).

The *fighting stance* strategy is essentially a single-step strategy, i.e. after only one step is taken, the posture is kept stationary to avoid further stepping. However, the execution of the *fighting stance* was different from pure stepping, as further compensatory steps were executed depending on the severity of the perturbation. The main difference compared to the pure step strategy was that emphasis was put on the execution of stepping to position the lower body into this specific *fighting stance*. This was executed through more effective and stable compensatory steps, characterised by a longer step length which lowered the CoM and increased stability. The lower body looked more stable through the front and rear leg, with flexed ankle, knee, and hip, and coupled with pelvic and trunk rotation. Some external rotation at the ankle of the rear leg was also observed, with more pronounced external rotation in some volunteers. Usually, compensatory steps were utilised to increase stability after finding the stance, to modify the stance width which increases the BoS. This was often executed using the rear leg, as the front leg acted as the weight-bearing component in the stance. In contrast, all pure steppers displayed multiple stepping with different execution than the *fighting stance* volunteers (less stable and no particular lowering of the CoM), with no clear display of finding a specific stance to stop the stepping, throughout all perturbation trials. In trials where pure steppers achieved 2 points, the final position, either a stance or returning to the starting position, looked differently compared to the *fighting stance*

(less stable with smaller step length) and such stationary position was not achieved for more severe perturbations.

With respect to the perturbation trials, the main recovery strategies were the *fighting stance* (10 out of 24) and the step strategy (12 out of 24). Two other strategies were identified but were only unique cases and has been left out in this report. Based on the highest scores, seven out of 13 males utilised the fighting stance, of which five scored 2 points for most of the pulses, covering the top five ranking out of all volunteers. Only three out of 11 females utilised the fighting stance, with two of them ranking at the top of the grading table, below the five most successful male users of the strategy.

The rearward-facing perturbations were the most severe conditions. The volunteers had higher success rates for balance during the *higher jerk* perturbations. In general, all volunteers expressed consistency in their preferred balance recovery strategy throughout all perturbations in both forward and rearward facing orientation. The results indicate that the *highest acceleration* was the most troublesome perturbation, with a slight disadvantage during the corresponding rearward-facing orientation (13% vs. 4% success). The highest rearward-facing braking pulse was the next troublesome perturbation (50% success). The forward-facing highest magnitude braking pulse had the highest success rate (83%). However, due to safety considerations for some volunteers, not all volunteers participated during the *highest braking* pulses as indicated by the lower number of tests performed (18 out of 24 volunteers). The number of females participating for that pulse was decreased from 11 to seven, and for the males from 13 to 11. The forward *highest jerk* and *lowest braking* pulse had the highest success rate including all 24 volunteers (79% success each), with the rearward *highest jerk* and *lowest braking* pulse having a similar success rate (74% vs. 71%). Overall, the females were more successful than the males in the rearward perturbations, and vice versa.

The *highest jerk* perturbation induced the fighting stance faster and more successfully to recover balance and adopt a stable stance. Furthermore, details from the video analyses show that the *lowest braking* was usually not severe enough to cause balancing instability for the majority of the volunteers. The volunteers' recovery strategies (mostly the fighting stance) were not challenged, and the execution was not problematic. In general, those who achieved 1 or 2 points for the *lowest braking* pulse, had little to no difficulties in balance recovery and at most exhibited only compensatory steps at the second half of the perturbation or utilised one step to find the stance (but with less displayed instability). Furthermore, the majority recovered balance fully, meaning that the volunteer returned to the original starting position on the force plate or remained in a stance position with minor body displacement.

2.4 Discussion

The identification extended the definition of the step strategy induced during more severe perturbations, while the characterisation provided descriptions of the execution of the strategies to illustrate how stepping responses differ. The main difference was the effective recovery steps displayed by the *fighting stance* volunteers compared to the pure steppers. Here, the *fighting stance* users utilised steps that were indicative of attempting to execute and finalise the stance quickly, lowering the CoM by increasing ankle, knee, and hip flexion together with trunk and pelvic rotation. If multiple steps were displayed, these were usually used (if balance recovery was achieved i.e. achieved 1 or 2 points) to correct the stance to find a more stable posture and to increase the BoS. The pure steppers never displayed such effective stepping, with no clear display of lowering the CoM and extending the BoS area, and no clear stability gained from each step as the single-support phase was very brief (i.e. quick contact-on and contact-off with each step).

The higher recovery rates for the *fighting stance* volunteers explain the effectiveness of this stance and accompanied stepping execution, as previous studies have shown that effective steps with increased step length of the front leg led to better balance recovery compared to smaller and multiple steps (King et al., 2005). The effect of withstanding multiple stepping, by using single steps, has been shown to increase balance recovery among the elderly (Carty et al., 2012). However, with increased perturbation magnitudes, multiple steps were also encountered among the *fighting stance* users. Previous studies have shown that multiple stepping is an indicator of fall risk (Carty et al., 2012; Maki & McIlroy, 2006). Although compensatory stepping in the *fighting stance* was effective in increasing postural balance and thus recovery,it was apparent that increased perturbation severity made it more difficult to execute such effective recovery steps for the purpose of finding the stance. The stance was harder to execute and maintain. Hence, when balance recovery was not achieved for the *fighting stance* users, their stepping behavior was more similar to pure steppers.

This can indicate two options observed in this study during recovery stepping, either: 1) pure steppers were not strong enough to execute a stance, or 2) pure steppers never intended to execute a stance and only tried to step towards recovery and balance equilibrium. Since the *fighting stance* users clearly displayed intention of executing a typical stance, whilst the pure steppers never displayed this characteristic, option 2) seems more likely. This needs quantitative identification in a biomechanical manner. From this qualitative study, pure steppers that achieved balance recovery never displayed execution of compensatory steps that indicated the goal of finding a stance position. This intention was apparent in the execution for all *fighting stance* volunteers. Here, the characterisation of the *fighting stance* with the lowered CoM, increased lower body joint flexion, rotations at the hip and trunk, provides insight and detail that differentiates a pure stepper from a stance user. However, this shows that although the *fighting stance* volunteers had a higher success rate with a more effective strategy, the perturbations in this study were too severe. Thus, the perturbation thresholds with respect to balance recovery among healthy younger people was found.

Overall, the higher jerk perturbations allowed the volunteers, especially the *fighting stance* users, to execute the recovery strategy and adapt to maintain balance through quicker initiation of the first recovery step. For the higher acceleration, the majority of the volunteers were unsuccessful in recovering balance, although a few *fighting stance* users were able to withstand the forward pulse. During the higher braking, the most successful volunteers recovering balance were clearly the *fighting stance* users. Hence, the effective *fighting stance* might have lowered the harness deployment rate. However, some volunteers, mainly pure steppers, were excluded from the trials due to safety considerations and hence this finding can be misleading without the responses of the excluded pure steppers.

Although the *fighting stance* might presumably require greater muscular control and engagement, comparing this to successful single-step strategies needs to be investigated as this has, to the authors' knowledge, not been characterised when differentiating a single-step strategy and multiple-step strategy in previous studies. Nevertheless, as single-step strategy is more effective than multiple-step strategy to recover balance, it is interesting to investigate if instructing the *fighting stance* with its defined characteristics, can improve balance for older people but also in public transport settings when hand bracing is unavailable. The take-home message of this study is that identification of different stepping strategies displays different characteristics of resulting recovery effectiveness. It is useful to provide details surrounding individual responses to perturbations, which allows perturbation thresholds with respect to postural balance to be determined. The findings of this study can also be utilised for future studies to provide details of volunteer responses which allows for differentiation of stepping strategies to be communicated in the literature. Understanding free-standing passengers' postural balance will be needed to develop a safe operation envelope, especially with the development and introduction of automated vehicles in public transport.

2.5 Conclusion

This qualitative study shows that identification and characterisation provide details on volunteer responses to different perturbations. It can provide insight on the effectiveness of different stepping strategies, which can be useful to instruct public transport passengers in case of risk of falling and also in perturbation-based training for e.g. older people. Overall, the volunteer responses were differentiated into two groups, *fighting stance* volunteers and pure steppers. The *fighting stance* volunteers executed more effective compensatory steps, as displayed by their recovery adaptations to more severe perturbations. The higher jerk perturbations evoked faster initiation of the first recovery steps leading to more successful recovery for most volunteers, especially for the *fighting stance* volunteers. These volunteers showed higher success rates overall and ranked at the top based on their effectiveness of balance recovery. The higher acceleration profiles were too severe and should not be reached in practice to reduce the risk of falling in free-standing scenarios on public transport. Future studies should provide more details on the volunteer responses, such as the identification and characterisation of specific volunteer strategies, as this can provide insight and transparency among studies of similar nature in the literature.

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3 A biomechanical quantification of an effective balance recovery strategy in free-standing females and males using OpenSim

Jia Cheng Xu, Ary P. Silvano, Arne Keller, Simon Krašna, Robert Thomson, Corina Klug, Astrid Linder

Here, the short communication to IRCOBI Europe 2021 is presented with smaller sections and no subchapters. The intention was to present ongoing work in a smaller format. The final paper will differ from the manuscript presented here due to the review process.

Introduction:
Free-standing public transport passengers are subjected to driving maneuvers that can cause postural instabilities, increasing the risk of falling. Balance strategies are the motion patterns observed in humans to counteract perturbations. The study in Chapter 2 identified different recovery strategies and provided characteristics to evaluate the effectiveness through a qualitative video analysis from the volunteer tests described in Chapter 1. The most effective strategy was the *fighting stance* strategy, a single-step strategy with postural characteristics such as a weight-bearing front leg and a supporting rear leg to increase stability by lowering the CoM and broaden the BoS to recover balance. Thus, the *fighting stance* facilitates balance recovery by incorporating a stance to withstand further stepping. As this strategy shows great potential in increasing postural balance in free-standing scenarios, further investigation of this strategy, to understand its effectiveness, is needed.

Therefore, the current study assessed the *fighting stance* by deriving inverse kinematics (IK) to obtain joint angle characteristics among female and male users of this strategy. The aim was to provide baseline biomechanical knowledge to understand its execution during different perturbations and relate it to the resulting balancing outcome derived in the study described in Chapter 2. This was done using OpenSim (Delp et al., 2007), an open-source platform commonly used for musculoskeletal modeling and simulation of movements.

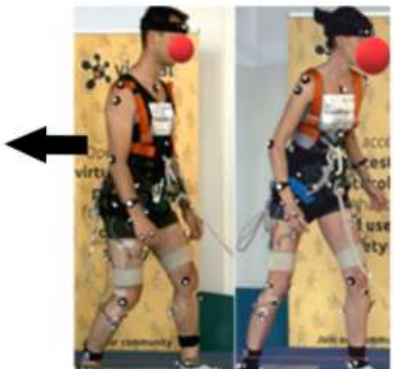

Figure 3.2. *Fighting stance* in rearward direction. Arrow indicates sled motion.

Method:
A full-body musculoskeletal model of an average male developed by Rajagopal et al. (2016), and denoted as the Rajagopal model, was used to perform scaling and IK on one female and one male volunteer. The marker coordinate data used were captured in the volunteer tests described in Chapter 1. Only responses from the rearward Acc1-J1 and Acc1-J2 were simulated, with Acc1-J1 acting as the baseline pulse and Acc1-J2 providing the higher jerk magnitude to understand the balancing response when jerk is increased. The ankle, knee, and hip angles were used to characterise the *fighting stance*.

Initial findings:

Joint angle responses for the ankle, knee, and hip for the female and the male volunteer were obtained from the OpenSim simulations. Figure 3.2 displays the balancing reaction where the *fighting stance* was executed. Positive slope of the curve indicates joint flexion and negative slope indicates joint extension. The main finding was the difference in joint angle response to the acceleration pulses, where *higher jerk* induced a faster joint angle response in the left leg for both volunteers. Compared to the *baseline* pulse, the joint angular response in *higher jerk* perturbations tended to a more constant value at the end of the pulse.

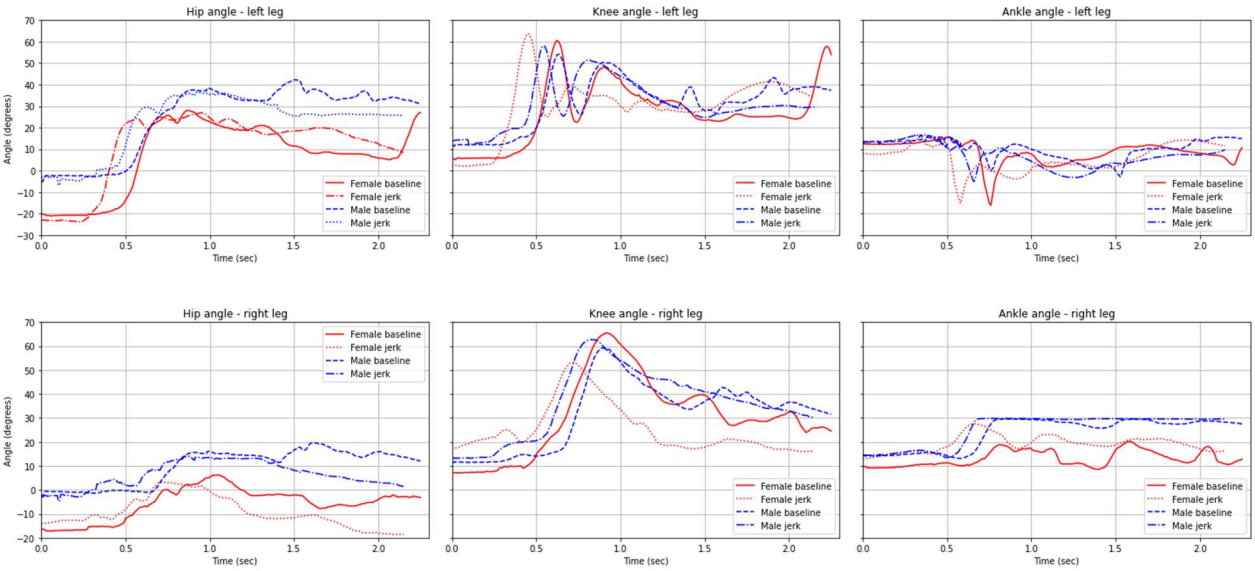

Figure 3.2. Joint angle responses for the ankle, knee, and hip for the female and the male volunteer.

Discussion:

The initial findings present the time history of the lower body joint angles relevant for characterising the *fighting stance*. Here, the focus was on the rearward-facing direction for two pulse severities of the *fighting stance*. Figure 3.2 displays hip and knee flexion of the front leg before a longer knee flexion of the rear leg, to position the left leg (Figure 3.1) in front of the body to initiate the *fighting stance*. The ankle acts to stabilise the foot to the ground, with the rear leg having a flexed knee and dorsi-flexed ankle (positive ankle angle), finalising the *fighting stance* execution. This characteristic should be noted when analysing the motion pattern of the *fighting stance*, as irregular joint kinematics (multiple peaks in the joint angle response) could indicate ineffective execution of the strategy which was seen for the *baseline* pulse compared to the *higher jerk*. This indicates that the *fighting stance* was executed more effectively in the *higher jerk* than *baseline* due to more stable motion patterns as illustrated towards the end of the perturbation, indicating balance recovery. Therefore, it is hypothesised, that the *higher jerk* might indicate a perturbation threshold that is enough to induce a strong initial reaction, provoking a more effective biomechanical response to counteract the perturbation. The synergies among the lower body joints, as well as further perturbation profile parameters, should be studied in more detail for the *fighting stance*. This might provide knowledge to assess driver maneuvers and interior designs for passenger safety in public transport.

From the study in Chapter 1 and 2, it was noted that higher acceleration and braking pulses had the lowest balancing success rate for all strategies, although the *fighting stance* was still the most

successful. Hence, the initial findings should be complemented with joint angle kinematics during different perturbation profile parameters, prioritising higher acceleration and braking perturbations. The effectiveness of the *fighting stance* can be further understood with more detailed muscle response modelling and utilising the biomechanical data recorded in the volunteer studies. Furthermore, the response of the volunteers applying the *fighting stanc*e should be compared to other balance strategies to assess the various strategies' effectiveness in terms of injury risk due to loss of balance. Biomechanical data including dynamics and muscle responses are needed for a more complete quantification. Eventually, active human body models (HBMs) should be developed due to their potential to assess injury risks using different balance strategies within different environments, and during various perturbations, for free-standing public transport passengers. Derived biomechanical data using musculoskeletal modelling softwares, such as OpenSim, can provide valuable inputs to HBMs.

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4 Accelerations of public transport vehicles: a method to derive representative generic pulses for passenger safety testing

Arne Keller, Simon Krašna

This is a short summary of a not yet submitted article and might differ from the final version.

With the goal of providing insights into the acceleration perturbations challenging the balance of standing passengers on public buses, we analyse acceleration time series recorded on buses in normal operation as well as on test drives. After splitting into acceleration and deceleration events, the pulses are studied using a Legendre polynomial expansion and a weighted mean in order to condense a larger set of events into representative average signals. As a result of the method presented, generic pulse shapes are derived, which can be used in the future as the basis of a standardised framework for volunteer experiments and virtual testing addressing the standing passenger problem.

4.1 Introduction

In the context of the risk of non-collision injuries of standing passengers, the acceleration and deceleration behaviour of the vehicles under consideration plays a crucial role as the actual perturbation signals challenges the postural balance of the persons at risk. Therefore, whether it is by experimental studies or numerical modelling, a deeper understanding of the postural balance of standing passengers and the related injury risks requires a solid knowledge on these underlying signals.

The acceleration behaviour of buses and trams has been in the focus of researchers and engineers since the early 20th century. However, until the 1970s, this work mostly had the goal to define comfort thresholds for acceleration magnitudes and jerks (Hoberock, 1977; Brooks et al., 1978), while not investigating the structure of the acceleration pulses more in detail. The first study that considered the influence of acceleration pulses on postural balance of standing passengers was done by Graaf and Van Weperen (1997), who measured acceleration time series on buses and trams and used similar perturbations in treadmill experiments with volunteer participants. More recently, several authors studied acceleration time series of buses (Zaworski et al., 2007) and subway trains (Powell and Palacín, 2015), with a focus on passenger (dis)comfort. The shape of average emergency braking pulses was investigated by Turkovich et al. (2011). Schubert et al. (2017) measured acceleration pulses during a volunteer study with a bus in test drive conditions – however, a closer analysis of the acceleration time series was not the main objective of their work.

The most systematic study investigating the structure and shape of acceleration pulses so far is the one by Kirchner et al. (2014). These authors expanded time-normalised acceleration/deceleration pulses in terms of Legendre polynomials. Furthermore, they suggested to select the event of highest mutual similarity of a given set as the most representative one for the set.

In spite of the research done so far, with a view towards providing suitable input data for computational and laboratory studies of the standing passenger problems, there are still significant gaps both in

knowledge and methodology. Not only are representative sets of field data still not available in the current literature – also a method is missing to automatically derive meaningful average acceleration/deceleration pulses from larger data sets.

In this work, we address this methodological gap by expanding the Kirchner et al. (2014) method to an automatically applicable algorithm, which could be used for datasets of much larger sizes. Furthermore, we propose a weighted-mean method to compute an average pulse from a given set of acceleration/deceleration pulses. We can show that this weighted mean represents, of all pulses representable in a Legendre expansion of given order, the one with the highest mean similarity with all pulses under consideration. As a demonstration of the method, we present a set of real-life acceleration data recorded on public buses during normal operation and on test drives, from which we derive generic acceleration and deceleration pulses representing the typical behaviour of the vehicles under consideration.

4.2 Data analysis method

4.2.1 Preprocessing and automatic splitting

The measured one-dimensional raw acceleration data are Butterworth filtered (second order, cutoff frequency 0.75 rad/s) in order to remove high-frequency oscillations. The resulting filtered acceleration signal is treated by an automated splitting algorithm based on cropping of constant phases and a discrete fourier transformation. As a result, we obtain a set of acceleration (ACC) and deceleration (DEC) events for each acceleration time series. For further treatment, these pulses are time-normalised to a dimensionless time variable 0 < t <1, resulting in a set of functions $f_m^{A/D}$ defined on the interval [0,1] describing the mth ACC and DEC pulses.

4.2.2 Legendre expansion

Expansions in Legendre polynomials are known as a handy tool e.g. in image processing (Paton, 1975). The mathematical properties of these expansions and their convergence are well-documented in the scientific literature – for an overview, see Wang and Xiang (2012). The idea to expand acceleration pulses in these polynomials originates from Kirchner et al. (2014). As opposed to the most common representation on the interval [−1,1], the shifted Legendre polynomials on the support [0,1] are used in this study.

Based on the orthogonality of the Legendre polynomials, the time-normalised functions resulting from the splitting algorithm can be approximated

$$f_m^{A/D}(x) \approx \sum_{k=0}^{N} \left(c_m^{A/D}\right)_k P_k(x),$$

where $P_k(x)$ is the k-th Legendre polynomial, $\left(c_m^{A/D}\right)_k$ the k-th Legendre coefficient of the m-th ACC or DEC event in the given set and the integer N is the order of the approximation. The Legendre coefficients can be determined either by integrating the orthogonality relation or by a least-square fit.

The Legendre approximation reduces the number of degrees of freedom for each pulse to a small number of coefficients (40-70 coefficients are mostly sufficient, except if jerks are to be estimated) and allows in a straightforward way to compare the shapes of different pulses (see subchapters 4.2.3 and 4.2.4).

4.2.3 Similarity analysis

The similarity coefficient of two normalised pulses is defined as the mutual correlation coefficient with respect to the square-integral scalar product. This coefficient can directly be computed from the

Legendre representation without additional numerical integrations. For a set of M (ACC ord DEC) pulses, this results in a symmetric M x M matrix containing M(M+ 1)/2 independent similarity coefficients. For each pulse of the set, the arithmetic mean of the corresponding row of the matrix is called mean similarity coefficient, which represents the mean similarity of this pulse with all the other ones of the set.

4.2.4 Averages/representative example: Maximum similarity and mean pulses

Similarity coefficient can be used to define a representative average or to choose a representative example for the shape characteristics of a set of time-normalised events without over-representing high magnitude events. Kirchner et al. (2014) suggested to select the pulse of highest mean similarity as a representative example – the so-called "maximum similarity pulse."

While the maximum similarity pulse method allows picking a representative example out of a set of pulses, in many applications (particularly when creating input to experimental or numerical tests), it is more appropriate to use an average that takes into account the shape of all pulses of the set to some extent. As a method to calculate such a representative average of the shapes of different pulses without over-representing high magnitude events, we suggest considering a weighted mean with the inverse of the L2-norm of each pulse as weight. It can be shown (*left out in this summary*) that the multiples of this pulse have, of all Legendre series of the same order, the highest possible mean similarity with the pulses of the set. We refer to these weighted-mean pulses as "mean pulses."

Due to the cutting algorithm and varying terrain gradients, the measured acceleration pulses tend to have an offset or to be lop-sided. Therefore, the mean pulses defined above in general do not equal to 0 at the start and end of the interval. It is possible, however, to slightly modify the method in that it gives the Legendre coefficients of the pulse that, of all Legendre representations of the same order that represent a function starting and ending at 0, have the highest possible mean similarity coefficients with all pulses of the given set. We refer to these as "constrained mean pulses."

4.3 Measurements and data

The first set of measurements have been carried out on several bus lines of the Zurich public transport network in normal operation conditions. The instrument used was a commercially available mobile phone (Samsung Galaxy S5) equipped with an application designed to read out the on-board sensors (three-axial accelerometer, gyroscope, GPS) every 0.02 seconds. The instrument was manually aligned with the vehicle and held inplace during travel.

A second set of measurements was performed on a closed track of 100 m length on dry tarmac. A city bus type MB Sprinter 616CDI (2007) with 12 seats and 17 places for standing passengers was used. During the tests, the driver and three staff members were on board. For the purpose of the study, the longitudinal acceleration profiles were measured during the bus accelerating and braking to full stop. The acceleration was measured with VC4000 vehicle performance computer (Veri-com, USA), located onboard at the centre line in the mid-distance between the front and the rear axles. The raw acceleration signals were low-pass filtered at 20 Hz cutoff frequency with a built-in analogue filter.

For both data sets, only the longitudinal acceleration data was taken into account for this study. In the following sections, the first data set is called "normal operation" (NO), while the second is referred to as "test drive" (TD).

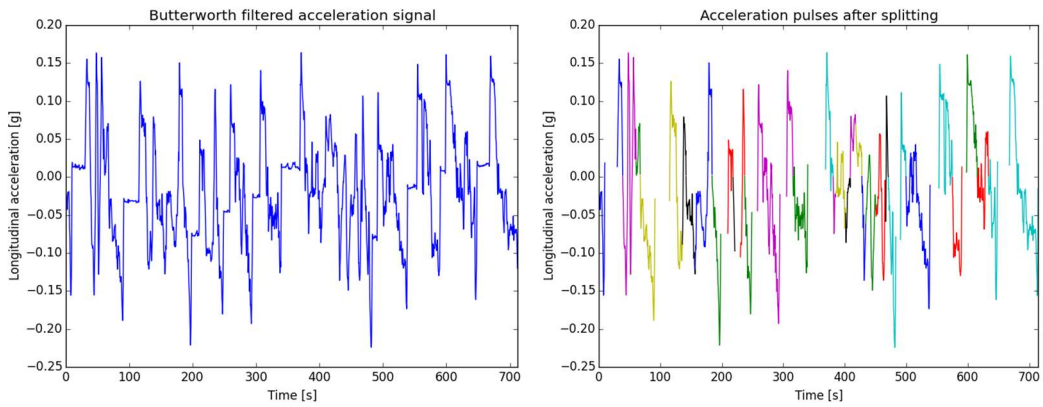

Figure 4.1: Example for splitting algorithm, NO data, electric vehicle.

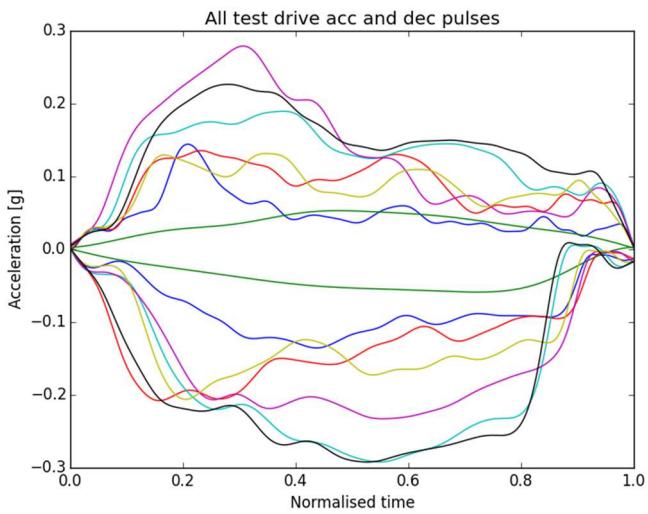

Figure 4.2: All TD ACC and DEC events.

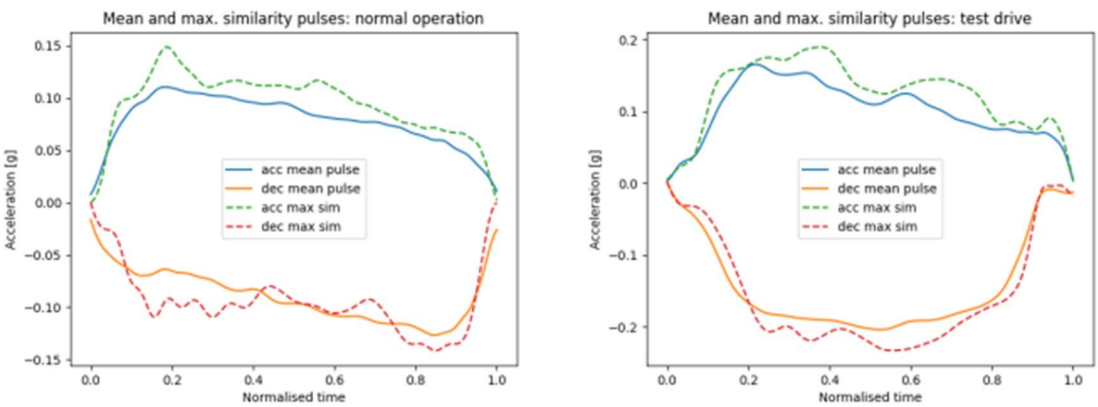

Figure 4.3: Mean and maximum similarity pulse, NO and TD

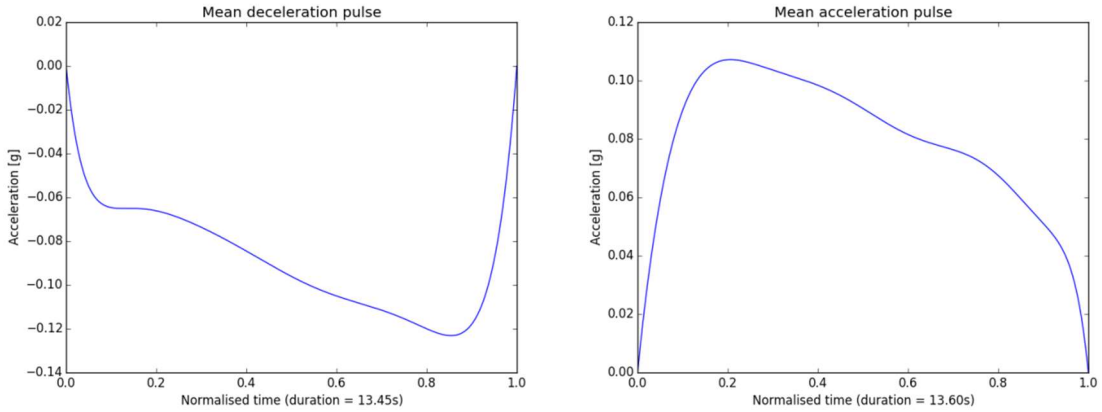

Figure 4.4: Constrained mean pulses for NO data, up to order N = 11.

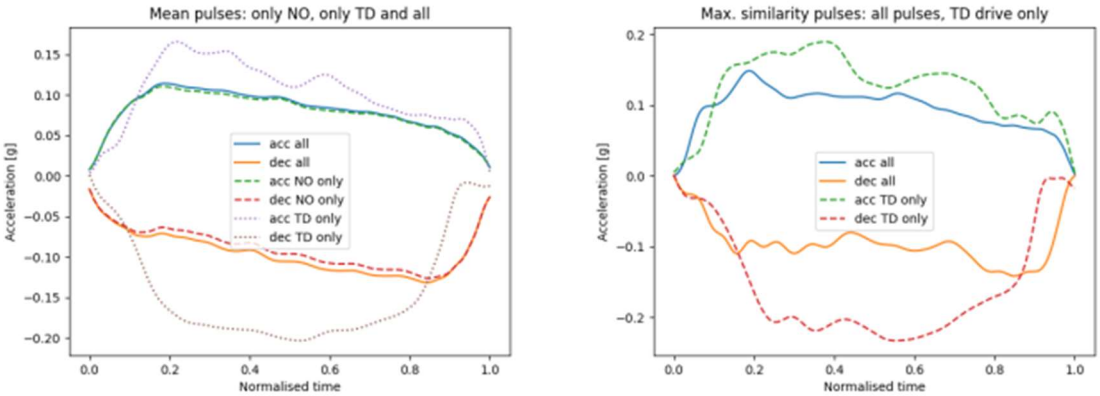

Figure 4.5: Comparison normal operation and test drive pulses. Left: highest mean similarity acceleration and deceleration pulses for only TD, only NO (which are also the highest mean sim. pulses for all data taken together). Right: mean acceleration and deceleration pulses, only NO, only TD, both.

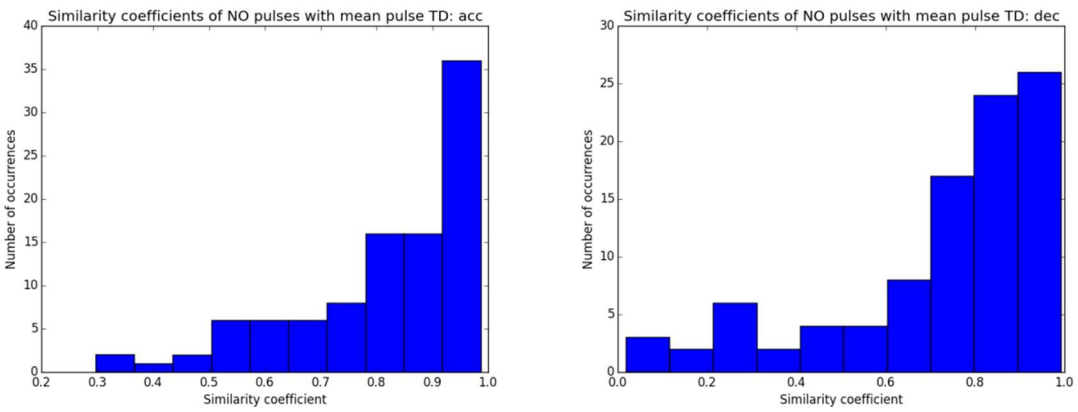

Figure 4.6: Test drive vs. normal operation: Histograms of similarity coefficients of mean TD pulses with all NO pulses, acceleration and deceleration.

Table 3: Similarity coefficients of the mean pulse (mp) and the maximum similarity.

Pulse set	Similarity coefficient mp vs. ms	
	acc	dec
all	0.9970	0.9848
NO	0.9958	0.9840
TD	0.9933	0.9953

4.4 Results

Each of the "normal operation" time series was split into acceleration and deceleration events according to the splitting algorithm. Altogether, the splitting of the normal operation data sets resulted in 99 ACC events with magnitudes up to 0.215 g and 97 DEC events up to −0.279g. As an example, the splitting of one of the time series (after Butterworth filtering) is shown in Fig. 4.1.

For all events under consideration, Legendre expansion up to order N= 10 (i.e., 11 coefficients) and N = 50 have been computed. All ACC and DEC events of the TD data are displayed in Fig. 4.2.

4.4.1 Mean pulse method, maximum similarity pulses

The weighted means of the NO events have been evaluated with the unconstrained and the constrained mean pulse method. The unconstrained mean pulse results, together with the maximum similarity pulses, are shown in Fig. 4.3 for both data sets. Furthermore, in Fig. 4.4, the constrained mean pulses for the NO data are given. Furthermore, in order to compare the results of the two methods directly, in Fig. 6 the maximum similarity and mean pulses for different sets are displayed.

4.4.2 Similarity coefficients

Sets of similarity coefficients are useful (i) to compare the test drive and normal operation results and (ii) to compare the performance of the mean pulse and maximum similarity pulse methods. For the first purpose, histograms of the mean similarity coefficients of the test drive ACC events with all normal operation ACC events (and the same for deceleration events) are displayed in Fig. 4.6. In order to facilitate the comparison of the results of the two different average pulse methods, the mean similarity coefficients of the results with different sets of pulses are given in Table 3. Furthermore, the direct similarity coefficients of the mean pulse and maximum similarity results are shown in Table 4.

4.5 Discussion

4.5.1 Splitting and Lagrange representation

As visible in Fig. 4.1, the splitting algorithm manages to consistently detect the bigger spikes in acceleration and deceleration – even though in some points, the trained eye would probably have subdivided some events into several ACC and DEC pulses in a manual splitting. This is also apparent from the unequal number of ACC and DEC pulses. On the other hand, the more regularly shaped events in the test drive data set are clearly recognised by the splitting algorithm. It is clear that a manual splitting, as done by Kirchner et al. (2014), would improve the quality of the pulse splitting in tricky cases. However, an automatic splitting is mandatory as soon as the data sets reach a certain size, so we consider it important to develop the splitting algorithm further.

Table 4: Mean similarity coefficients over different subsets of the average pulses according to maximum similarity (ms) and mean pulse (mp) method, based on different subsets

Method, direction and basis set of average pulse	Mean similarity coefficient over subset		
	all	TD	NO
mp acc all	0.835	0.962	0.826
mp acc TD	0.825	0.973	0.815
mp acc NO	0.835	0.960	0.826
ms acc all	0.831	0.961	0.823
ms acc TD	0.824	0.961	0.814
ms acc NO	0.832	0.961	0.821
mp dec all	0.799	0.915	0.790
mp dec TD	0.745	0.981	0.728
mp dec NO	0.798	0.903	0.791
ms dec all	0.787	0.904	0.778
ms dec TD	0.742	0.973	0.725
ms dec NO	0.787	0.904	0.776

4.5.2 Comparison normal operation and test drives

While a deeper analysis of the pulse shapes is beyond the scope of the present work and data sets, it is still interesting to compare the overall appearance of the mean pulses obtained in both data sets. While the acceleration pulses both show a fairly similar behaviour with a strong rise in the beginning, followed by a slower decrease and a drop in the end, the deceleration pulses differ substantially in shape (Fig. 4.5), where the normal operation data is rather characterised by a gentle slope and a drop in the end, while the test drive data show much larger magnitudes (even in the mean pulse) with strong rise and drop in the end. This difference of the acceleration and deceleration pulses in normal operation and test drive is also visible in the similarity coefficients of the mean and maximum similarity pulses with different parts of the dataset (Table 3, Fig. 4.6), where the TD deceleration pulses show lower similarities with the normal operation ones, both in mean and in the histograms (while still showing mean similarities > 0.7). Taking into account that the conditions under which the two data sets were obtained were substantially different (a bus driver with passengers compared to a test driver), these differences are not surprising and particularly the asymmetry acceleration – deceleration could be due to the more powerful emergency braking capabilities of the vehicle. Nevertheless, there seems to be a typical acceleration and (to a lesser extent) deceleration behaviour which can be observed similarly in different vehicle types, which is interesting to study with respect to its impact on passenger.

4.5.3 Maximum similarity and mean pulses

Both the maximum similarity method and the mean pulse method aim at extracting a representative example or average from a number of events with different magnitudes without being dominated by the events with the highest magnitude. When looking at the resulting pulses based on the normal operation data, it is apparent that both methods manage to capture the typical features of acceleration and deceleration pulses (see above). Also the constrained method reproduces these properties. The direct comparison of the mean pulse and maximum similarity results (Fig. 4.3) shows that the results are surprisingly similar; an observation which is confirmed by mutual similarity coefficients all>0.993 for the mean pulse and the maximum similarity results for the different pulse sets. The most apparent difference between the results of the two methods is that the maximum similarity pulses show the spikes and random oscillations which are typical for measured acceleration events, while in the mean

pulse results these disappear in average over the whole set. This average property of the mean pulse method is also visible when the comparing the results obtained on different subsets (only NO, only TD, all) in Fig. 4.5. While the results of the test drives, particular in deceleration, differ substantially from the normal operation mean pulses, both in shape and in magnitude, the mean pulse taking into account all data only changes very little compared to the mean pulse on the normal operation data set (which is much bigger than the test drive set). The maximum similarity results, on the other hand, are also substantially different for the NO and TD data sets, while the result on the combined data set agrees exactly with the NO result (as the maximum similarity pulse happens to be part of the latter). Therefore, the maximum similarity method does not only deliver more random results, but also lacks a regular average behaviour when applied to different data sets.

4.6 Conclusion

In this study, a novel method has been presented to define representative average shapes of acceleration and deceleration pulses of public transport vehicles, based on measured acceleration time series. The method consists of a combination of automated splitting, Legendre polynomial fits, similarity coefficients and a weighted-mean approach to capture the average shapes without being dominated by events of large magnitude. In order to test the method and to present some pulse shapes as first results, two data sets measured on buses in normal operation and on test drives were considered, allowing a comparison of the new mean pulse method and the (established) maximum similarity method. The results show that the method is capable of automatically extracting meaningful representative averages out of the data sets, which due to the weighting are not dominated by the events of the strongest magnitude. Due to the average properties of the method, the mean pulses are free of random oscillations typically occurring when picking a representative example out of the set, thus being suited ideally for the application as input to future laboratory and numerical studies of standing passenger safety. Furthermore, with the automatic splitting algorithm, the approach has the potential to also be applied to larger real-world data series.

4.7 Acknowledgement

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