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What is This?

Investigating the injury risk in frontal impacts of Formula Student cars: a computer-aided engineering analysis

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Abstract: This paper presents a parametric study of the injury risk in collisions of race cars designed for the European Formula Student competition. The study is motivated by the fact that only a limited assessment of driver safety is required for this competition. The approach was to model a Formula Student car in a mathematical dynamic model environment. A parametric study was then carried out to investigate the sensitivity of injury to various system variables. These were the crash pulse, the occupant stature, and the occupant posture. These system variables, under close examination, can be changed to alter the occupant kinematics or, in other words, they change the injury risk. The results of the analysis showed that the risk of injury in a frontal impact was dependent on the system variables. The risk of an abbreviated injury scale (AIS)2+ injury was 22.3 per cent in the baseline constant-g test, increasing to 35.2 per cent in the worst case. For AIS3+ the values were 5.1 per cent and 11 per cent, respectively. The study also showed that the occupant restraint conditions in a Formula Student car had a significant influence on the distribution of the injury risk between the body regions. The variation in the injury risk highlighted by this study, both in absolute terms and in the distribution between the body regions, showed that there are limitations to the use of vehicle kinematics in their current guise as a predictive tool for the injury risk. The results of this study represent a significant step in the understanding of the injury risk in a Formula Student frontal impact.

Keywords: Formula Student, Formula SAE, injury criteria, frontal impact, impact attenuator, mathematical dynamic model, crash pulse

1   INTRODUCTION

Formula Student is Europe's largest Student motorsport event and challenges young engineers to design and build a single-seater race car from scratch. The cars are then put through their paces in a series of dynamic events including acceleration, skidpan, and endurance. To ensure the safety of competitors, regulations provide a set of standards for the safety of drivers in front, side, and roll-over crash events. In

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developing these standards the philosophy has been to ensure that, for front, side, and roll-over crashes, the driver is enclosed within a strong survival cell and that, for frontal impact, there is a limit to the acceleration acting on the human body. The injury risk is not directly assessed by the regulations.

This study was motivated by the fact that only a limited assessment of driver safety is required for the Formula Student competition. The objective of the research was to investigate the injury risk in a frontal impact, this impact being the more highly regulated crash event requiring both a strong survival cell and a limit on acceleration. The requirement was to understand the link between the vehicle kinematics and the occupant injury. An

understanding of the occupant injury would represent a significant step in the process of evaluating and improving Formula Student safety.

2 BACKGROUND

In the 1960s, one out of every eight accidents in Formula One resulted in either a fatality or a serious injury [1]. While safety rules have been written over the years in reaction to injury-producing incidents, it is perhaps surprising that there has been a poor history of statistical data gathering in support of this activity [2]. The step change was the introduction of the accident data recorders (ADRs). Initially introduced in the US Indy Car series in 1993, these devices quickly moved into Formula One (introduced in 1997) and have since permeated into other motorsport formulae [3]. The device resembles the black box in an aircraft and records all the speed and deceleration data, which provide the basis for further safety improvements [4].

The use of ADRs has supported the science and testing that has been carried out in understanding the injury tolerance of occupants to severe impacts. The first systematic analysis of ADR data was by Melvin *et al.* [5] in 1998. The study showed that, for a restrained occupant, protection can be achieved in frontal, side, and rear crashes with severities in the peak deceleration range of 100–135 g and velocity change range of 50–70 mile/h. More recently, follow-up papers on injuries in US racing have tended to concentrate on the effects of the head and neck support device, for which statistical data are still being collected [2]. The deceleration data derived from ADRs have also been used to support the development of vehicle and circuit safety standards. An example is the Circuit and Safety Analysis System of the Fédération Internationale de l'Automobile [6]. An ADR has also been used as a predictive tool for injury risk. A severity index coefficient has been developed to enable impacts to be assessed quickly for their potential to cause head injuries, in order to rate them according to their need for fuller analysis [6]. The Indy Racing League also used the crash recorder to assess crash severity and to turn on a light that alerts medical and safety workers to a high-g impact having the potential to produce injury [3].

The availability of ADRs has been instrumental in the development of regulations that use vehicle kinematics as a surrogate for injury risk. Formulae as diverse as Formula One and Formula Student now seek to control acceleration–time histories during an impact in an effort to limit the occupant injury risk. Such an approach is not new. Since 1981, the US procedures for the assessment of roadside restraint systems [7] have related the occupant injury risk to the vehicle kinematics using the flail space model (FSM). The European procedures [8] use a variation of the FSM in conjunction with the acceleration severity index (ASI) to gauge the occupant injury risk. However, recent studies have highlighted limitations with both the FSM and the ASI as predictive tools for the occupant injury risk. Gabauer and Gabler [9, 10] reported that occupant injury assessment models based on vehicle kinematics are not always able to predict the occupant risk for all occupant restraint conditions.

To support the imposition of vehicle kinematics as a surrogate for the injury risk the influence of the restraint system needs to be well understood. In motorsport impacts, the near-direct coupling of the driver's torso to the chassis allows direct inference of the loads on the torso. However, the extremities are not closely coupled to the chassis of the car in the same way as the torso is. For example, the torso restraints only provide control of the corresponding crash motion of the driver's helmeted head by holding the shoulders and relying upon the neck to restrain the head. This significantly increases the risk of neck injury in severe impacts [11]. Laboratory and computer modelling has been undertaken in order to obtain specific information on the loads to body regions such as the head and lower extremities in motorsport impacts. Weerappuli *et al.* [12] developed a computer model to help to predict the injury potential to race drivers during crash events. Begeman and Melvin [13] used mathematical modelling of driver kinematics in instrumented racing car crashes to estimate the driver responses and loading in severe rear, side, and frontal impacts.

The introduction of ADRs and associated computer modelling has supported the development of regulations that seek to minimize the risk of injury in high-end motorsport formulae such as Formula One and the Indy Racing League. The efficacy of using vehicle kinematics as a surrogate for the injury risk now needs to be explored for Formula Student. The regulations for Formula Student require the recording of acceleration–time histories for a front barrier impact at 7 m/s. The use of these data in computer modelling to obtain specific information on the loads to body regions represents one step in the process of evaluating and improving Formula Student safety.

3 METHODS

The approach was to develop a model of a Formula Student car in a mathematical dynamic model

(MADYMO) environment and to use this model as part of a parametric study to investigate the injury risk. MADYMO is a leading multi-body dynamics solver and frequently used for automobile occupant safety and injury calculations. The following sections discuss the model configuration, the selection of assessment criteria, the model validation, and the selection of parameters for the parametric study.

3.1 Model configuration

The Cardiff University Formula Student car was modelled within the MADYMO crash environment. The expected level of analysis corresponded to an early design phase. Accordingly, the Formula Student car was described by a few global design parameters. The advantage of such a formulation is that it allows a good understanding of the system behaviour and is convenient for conceptual improvements. The individual elements of the system were as described below.

3.1.1 Vehicle exterior

A non-deformable model of the passenger compartment was developed. The passenger compartment was considered rigid in order to have a computationally efficient model for conducting the parametric study.

3.1.2 Vehicle interior

The steering wheel and column were modelled using ellipsoids. The steering wheel was connected to the passenger compartment using revolute joints. The brake and accelerator pedals were modelled using ellipsoids. Revolute joints allowed rotation of the pedals around the fixing points.

3.1.3 Restraint system

The harness was modelled in MADYMO by means of a hybrid belt system. This system consisted of a multi-body belt (attached to the vehicle anchor points) and a finite element belt to define the contact with the occupant. The lap portion was connected to a belt between the legs and there were two shoulder belts, making a total of five points of attachment to the seat. The lap portion was modelled as an ellipsoid.

3.1.4 Occupant

For this study, the Hybrid III anthropomorphic test device (ATD) model from the MADYMO database was used to simulate the occupant. An ATD is a mechanical model of the human body that is used as a human surrogate in crash testing. The model ATD is representative of the physical ATD. The model has the same basic geometry, inertial properties, joints, and stiffness functions. The model is represented in the MADYMO environment by rigid ellipsoid bodies connected by kinematic joints. Forces and moments are recorded at the same positions as for the physical ATD. A head-supported mass was included in the simulations to represent the crash helmet used in Formula Student. The choice of the ellipsoid dummy was based on the efficient run time and was therefore suitable for a parameter variation study.

3.1.5 Kinematics

The model is driven by acceleration pulses that approximate the crash profiles. The approach was to use crash profiles from actual Formula Student impact attenuator designs. A number of Formula Student teams were approached to provide crash profiles and information about the design of their impact attenuator. In selecting crash-pulse curves for use in this study the criterion was to include a wide variety of design solutions and crash-pulse curve shapes.

3.2 Injury criteria

In a crash event, physical injury will take place if the biomechanical response is of such a nature that the biological system deforms beyond a tolerable limit resulting in damage to anatomical structures and/or alteration in normal function. Mechanical surrogates of humans, rather than living humans, are used in crash tests to evaluate the safety attributes of vehicles. These surrogates, more commonly referred to as ATDs, measure engineering variables, such as forces, velocities, deflections, and accelerations.

To determine the injury risk from a crash test the measurements from the ATD must be translated to the risk of injury. For decades, work has been performed on human injury from blunt trauma in the automobile field. Simulated automobile crashes and/or impact tests are performed (replications of impacts in which the injury outcome is known or can be approximated), and the response of the biofidelic surrogate (cadavers, porcine subjects, etc.) is taken to represent the response of a human in that crash scenario. This response may be used in the development of numerical relationships between measurable engineering parameters and the risk of injury for that crash scenario.

The process of measurement, calculation of the injury criterion (the physical parameter, or function

of several physical parameters, which correlate well with the injury severity of that body region), and determination of the injury risk will be discussed by body region.

3.2.1 Head

The currently used worldwide regulatory criterion for controlling head injuries is commonly known as the head injury criterion (HIC). The HIC is calculated from the acceleration observed at the centre of mass of the head of an ATD. It is expressed as [14]

$$HIC = \sup_{t_1, t_2} \left[\left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a \, dt \right)^{2.5} (t_2 - t_1) \right]$$

where a is the acceleration expressed as a multiple of the acceleration due to gravity, and t_1 and t_2 are any two points in time separated by not more than 15 ms or 36 ms dependent on the regulation.

The expanded Prasad–Mertz curves [15] were used to determine the probability of head injury based on HIC 15 ms values. The curves were developed on the basis of the data from cadaver experiments in which the relationship between the HIC score, the skull fracture, and the brain damage were observed. Six risk curves were available, each corresponding to a level on the abbreviated injury scale (AIS) (a measure of the threat to life in impact-induced trauma [16]). The curves relating to AIS2 + (moderate to fatal) and AIS3 + (serious to fatal) were deemed to provide a suitable starting point for the purpose of this study. An AIS2 + head injury would start at unconsciousness for less than 1 h and an AIS3 + injury would start at unconsciousness for 1–6 h [17]. The equation for the AIS2 + and AIS3 + injury risk curves were [18]

$$AIS2 + : P = \frac{1}{1 + \exp^{(2.49 + 200/HIC - 0.00483HIC)}} \quad (2)$$

$$AIS3 + : P = \frac{1}{1 + \exp^{(3.39 + 200/HIC - 0.00372HIC)}} \quad (3)$$

where AIS2 + :P is the probability of a moderate to fatal head injury, AIS3 + :P is the probability of a serious to fatal head injury, and HIC is the head injury criterion (15 ms).

3.2.2 Neck

The neck injury risk was evaluated on the basis of N_{ij}. In the term N_{ij}, ij stands for the four modes of loading the neck: ij = TE denotes the tension–extension mode N_{TE}; ij = TF denotes the tension–flexion

mode N_{TF}; ij = CE denoted the compression–extension mode N_{CE}; ij = CF denotes the compression–flexion mode N_{CF}. The N_{ij} value is the sum of normalized loads and normalized bending moments according to [14]

$$N_{ij} = \frac{F_z}{F_{zc}} + \frac{M_{0Cy}}{M_{yc}}$$

where F_z is the force at the transition from the head to the neck, F_{zc} is the critical force, M_{0Cy} is the total moment, and M_{yc} is the critical moment.

The injury risk curves for N_{ij} have been developed using experimental data from porcine subjects. The load configuration on which the curves were based was tension–extension and hence N_{TE} was the value used to assess the injury risk. The injury risk equations for AIS2 + and AIS3 + were [18]

$$AIS2 + : P = \frac{1}{1 + \exp^{(2.054 - 1.195N_{ij})}} \quad (5)$$

$$AIS3 + : P = \frac{1}{1 + \exp^{(3.227 - 1.969N_{ij})}} \quad (6)$$

where AIS2 + :P is the probability of a moderate to fatal neck injury, AIS3 + :P is the probability of a serious to fatal neck injury, and N_{ij} is the neck injury criterion.

3.2.3 Chest

Chest injury can be evaluated on the basis of the sternum deflection, the sternum deflection rate, the viscous criterion, and the thoracic spine acceleration. Chest compression was chosen for the purpose of this study as it had been found to be a superior indicator of chest injury severity [19]. The injury risk equations for AIS2 + and AIS3 + were used in this study. These were [18]

$$AIS2 + : P = \frac{1}{1 + \exp^{(1.8706 - 0.04439D_{max})}} \quad (7)$$

$$AIS3 + : P = \frac{1}{1 + \exp^{(3.7124 - 0.0475D_{max})}} \quad (8)$$

where AIS2 + :P is the probability of a moderate to fatal chest injury, AIS3 + :P is the probability of a serious to fatal chest injury, and D_{max} is the maximum chest deflection (in mm).

3.2.4 Lower extremities

Risk curves for the knee–thigh–hip (KTH) injuries were developed by Kuppa *et al.* [20] and based on

the analysis of test data (126 single impact tests using whole cadaveric subjects) reported by Morgan *et al.* [21]. The results of the analysis suggested that the femur axial force alone was a reasonably good predictor of KTH injuries. The AIS2+ and AIS3+ risk curves were

$$\text{AIS2+}: P = \frac{1}{1 + \exp^{(5.795 - 0.5196F)}} \tag{9}$$

$$\text{AIS3+}: P = \frac{1}{1 + \exp^{(4.9795 - 0.326F)}} \tag{10}$$

where $\text{AIS2+}:P$ is the probability of a moderate to fatal KTH injury, $\text{AIS3+}:P$ is the probability of a serious to fatal KTH injury, and F is the maximum femur force (in kN).

3.3 Model validation

To validate the model, the occupant injury criteria obtained from simulation were compared with those obtained from physical tests. Kettering University conducted a number of tests at their Crash Safety Centre [22]. The tests were conducted at velocities of between 11 m/s and 13 m/s and used three crash-pulse shapes The occupant was a 50th-percentile Hybrid III ATD.

The validation was based on a comparison of the Kettering University constant *g*-pulse test with a comparative simulation performed in MADYMO. Two such tests were performed by Kettering University. For each of the response parameters, the peak values were obtained from the event time histories by Kettering University. The peak values were then normalized with respect to the each parameter's injury assessment reference value (IARV). IARVs represent the limits of the acceptable force, moment, etc., for each response parameter. The IARVs used in this study are shown in Table 1 [23].

A MADYMO simulation corresponding to the input parameters of the Kettering University tests was conducted (at an impact velocity of 13 m/s). Table 2 shows the results of the tests and the MADYMO simulation. The variation is based on the worst-case comparison of tests with the simulation. Only a partial validation was possible as the chest

compression was unavailable for the tests. However, for the three available parameters the simulation and test data compare well, showing a maximum difference of 10.7 per cent of the IARV for the HIC, only 4 per cent of the IARV for N_{TE}, and 9.7 per cent of the IARV for the femur axial force. The model was therefore considered to be suitable for a parametric study.

3.4 Parametric study

A parametric study was then carried out to investigate the sensitivity of injury to various system variables. The parametric study was based on exploring the sensitivity of the injury risk to changes in the system parameters. The system parameters were defined as variables that would be expected to alter occupant kinematics in a frontal crash event, were able to be altered within the bounds of the regulations, and would reasonably be expected to be different when considering Formula Student car design and use. The three system parameters identified were the crash-pulse shape, the occupant position, and the occupant stature. These will be discussed separately and a range of values identified for each.

3.4.1 The crash pulse

The dependence of the occupant response on the crash pulse has been well studied [24, 25]. In relation to the crash pulse, the Formula Student regulations stipulate the following [26]: '... the impact attenuator, when mounted on the front of a vehicle with a total mass of 300 kg and run into a solid, non-yielding impact barrier with a velocity at impact of 7.0 m/s, would give an average deceleration of the vehicle not to exceed 20*g*, with a peak deceleration less than or equal to 40*g* ...'

The regulation is purposely framed to allow the Student designer flexibility in their approach to the design of the impact attenuator. However, in doing so, while the regulation fixes the area beneath the crash-pulse curve (the energy at impact), the peak, and the average, the final shape is left open. Examples of basic crash-pulse shapes possible

Table 1 IARVs for the 50th-percentile Hybrid III ATD

Response parameter	IARV
HIC	700 (15 ms)
Neck injury criterion	1.0
Chest deflection	50 mm
Femur force	9.1 kN

Table 2 Test and simulation results for a constant-*g* crash pulse

	HIC 15 ms	N_{TE}	Femur force (kN)
MADYMO	231	0.47	4.7
Test 1	281	0.43	3.9
Test 2	306	0.50	4.9
Difference (% of IARV)	10.7	4.0	9.7

within the regulation include the following: linear increasing acceleration with time; constant acceleration; and linear decreasing acceleration with time. Each has been shown to result in different occupant kinematics (and hence different injury risks) during a passenger car crash event [25].

To understand how Student design choice can influence the injury risk, the approach was to use crash-pulse curves from actual Formula Student impact attenuator designs. A number of Formula Student teams were approached to provide crash-pulse curves and information about the design of their impact attenuator. In selecting crash-pulse curves for use in this study the criteria were to include a wide variety of design solutions and crash-pulse curve shapes. The final selection was as follows (Table 3). The acceleration–time histories for each crash pulse can be found by referring to Appendix 1.

3.4.2 The occupant position

Changing the angle of the seat back changes the posture of the occupant. While there is no rule within the Formula Student regulations relating to how far back the seat can be reclined, the rules relating to the lap belt mounting specify that, in side view, the lap belt for a reclined seating position (i.e. greater than 30° from the vertical) must be at an angle of between 60° and 80° from the horizontal [26]. Taking the angle of the lap belt from the seat back as 60°, this permits a maximum permissible (if not necessarily practical) seat back angle of 50° from the vertical. The seat angles used in this study were therefore 30° (the maximum upright position), 40°, and 50° (the maximum reclined position). Pictorial representations of the three seat angles are shown

in Fig. 1. For comparison, the typical posture in a high-level single-seater Formula car is typically in the region of 45° reclined. For a passenger car a value of 25° reclined is adopted for many regulatory testing requirements, e.g. those of the Insurance Institute for Highway Safety and the Research Council for Automobile Repairs, and also the Economic Commission for Europe Regulation 17.

3.4.3 The occupant stature

The Formula Student regulations require that the car must accommodate drivers whose stature ranges from 5th-percentile female to 95th-percentile male [26]. Given the vastly differing physical proportions, it is conceivable that they will have different risks of injury in a vehicle crash relative to their differing physical proportions. To investigate this, the following three sizes of ATD were used in this study: 5th-percentile female; 50th-percentile male; 95th-percentile adult male. These are shown in Fig. 2.

The injury risk curves developed for various injuries were taken to represent the risk of injury for a 50th-percentile male. These are applied and translated to dummies of other sizes through a process known as scaling. Scaling factors were obtained from the literature [18, 20] to determine the injury risk for a 5th-percentile female and a 95th-percentile male. These are shown in Table 4.

4 RESULTS

For each size of ATD a total of 15 simulations was performed, each simulation conforming to a unique combination of the crash pulse and the seat back angle. The full results set is available in Appendix 2.

Table 3 The impact attenuator design and crash pulse

Entrant*	Design	Crash pulse
University of Applied Science Esslingen	Aluminium foam	Constant acceleration
University of Applied Science Esslingen	Aluminium plate (fabricated box)	Early peak followed by period of constant acceleration
Technical University Dresden	Fibre-reinforced plastic (moulded)	Late peak followed by period of constant acceleration
Polytechnic Torino	Aluminium plate (fabricated box)	Variable (increasing acceleration)
ISAT (Nevers) Formula Student	Aluminium honeycomb (two layers)	Early and late peak

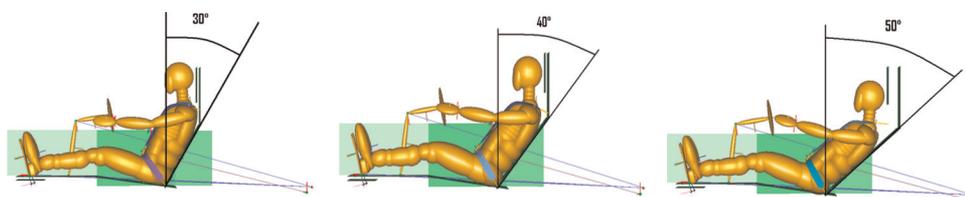*See Fig. 11 in Appendix 1 for details.

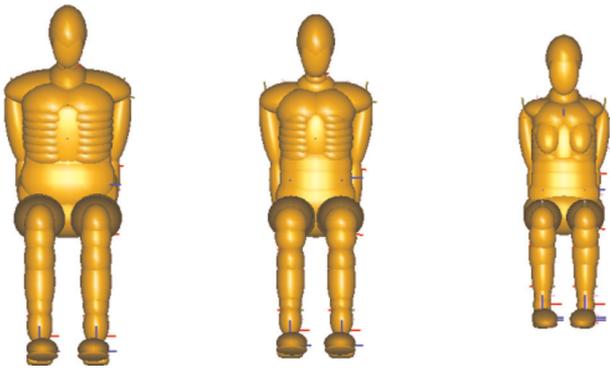**Fig. 1** 50th-percentile male with seat angles of 30°, 40°, and 50°

Fig. 2 95th-percentile male, 50th-percentile male, and 5th-percentile female

Table 4 Scale factors for the 5th-percentile female and the 95th-percentile male

| | Value for the following | |
	5th-percentile female	95th-percentile male
Head	1.0	1.0
Neck	1.0	1.0
Chest	0.83	1.11
Femur	0.72	1.17

A summary of the results is shown in Fig. 3 for AIS2+ injuries and in Fig. 4 for AIS3+ injuries. The baseline test was taken as the 50th-percentile ATD, constant-*g* crash pulse, and a 40° seat back angle.

From Fig. 3, it can be seen that the highest risk of an AIS2+ injury was for the chest followed by the neck and the femur. Changes in the input parameters have a significant effect on the chest injury risk. The injury risk increased from 22.3 per cent in the baseline test to 35.2 per cent in the worst case. On the other hand, the risk of a head injury was only slight and the change in the injury risk with input parameters was only 1.1 per cent.

For the AIS3+ injury, the neck takes precedence over the chest injury. The risk of a neck injury was observed to increase from 5.1 per cent (baseline and best case) to 11.0 per cent (worst case) with changes in the input parameters. The difference between the injury risk for the chest, neck, and femur body regions was less pronounced than for the AIS2+ injury risk. Again, the risk of a head injury was only slight and the change in the injury risk with input parameters was only 0.4 per cent.

5   DISCUSSION

The risk of injury, AIS2+, or AIS3+ is clearly dependent on the parameters under investigation. The
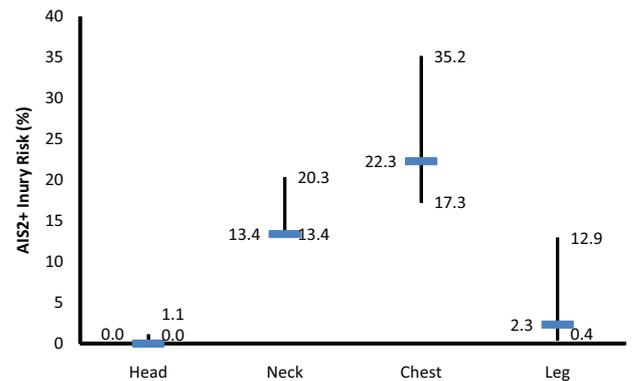

Fig. 3 AIS2+ injury risk by body region showing the baseline, the maximum, and the minimum values
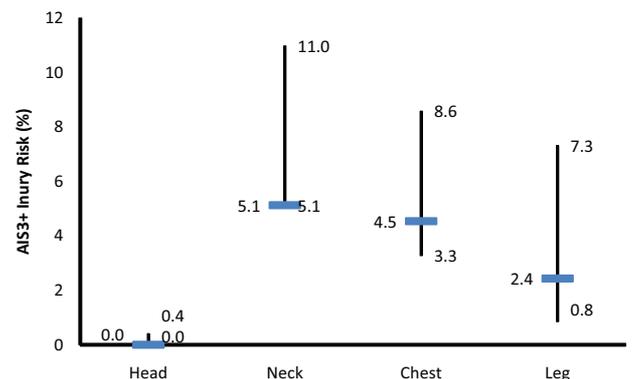

Fig. 4 AIS3+ injury risk by body region showing the baseline, the maximum, and the minimum values

baseline simulation predicted a maximum chance of an AIS2+ injury as 22.3 per cent and the maximum chance of an AIS3+ injury as 5.1 per cent. However, by altering the posture, the stature, and/or the crash pulse the risks of an AIS2+ injury or an AIS3+ injury was increased to 35.2 per cent or 11.0 per cent, respectively. This represents a significant increase in the risk of an AIS2+ injury or AIS3+ injury. In all cases the baseline configuration resulted in a risk of injury that was comparable with or close to the minimum expected injury risk for each of the body regions.

The analysis of means (ANOM) was the basis for an investigation of the change in the injury risk with changes in the crash pulse, the occupant stature, and the occupant posture. A baseline was established by averaging the response over the simulations in which the input parameter was the same as the baseline line test (e.g. for each simulation in which the occupant was a 50th-percentile male). Comparative measures were then established by averaging the response over the simulations in

which the input parameter was different from the baseline (e.g. for each test in which the occupant was a 5th-percentile female). The following sections discuss the influence of the three parameters on the injury risk. As the AIS2 + and AIS3 + results sets demonstrate similar patterns, only the AIS2 + results were discussed (the AIS2 + results showing the greater change in the injury risk). The ANOM results for the AIS3 + data set are shown for completeness.

5.1 The crash pulse

The ANOM for the change in the crash pulse is shown in Fig. 5 for the AIS2 + injury risk and in Fig. 6 for the AIS3 + injury risk. The baseline was the constant-*g* crash pulse.

With reference to Fig. 3 it is seen that the head injury was not a significant factor in the risk of an AIS2 + injury. For the HIC, the change in the crash pulse resulted in a negligible increase in the risk of an AIS2 + head injury. The analysis of the test simulations showed that the head did not impact any rigid structures and the acceleration of the head (the parameter upon which the HIC is based) was due to the restraint of the head through its attachment to the neck.

The risk of an AIS2 + neck injury was shown to be between 5.1 per cent and 10 per cent (Fig. 3). The motion of the head is a determinate in the neck injury criterion value; an impact of the head with a solid object would change the motion of the head and the modes of loading the neck. In these impacts the driver was closely coupled to the chassis of the vehicle and the head was prevented from impacting any solid objects. This resulted in similar motions for the heads in all the impacts and therefore only a negligible difference (less than 1 per cent increase) in the neck injury criterion. The change in the crash pulse was therefore not a significant factor in the
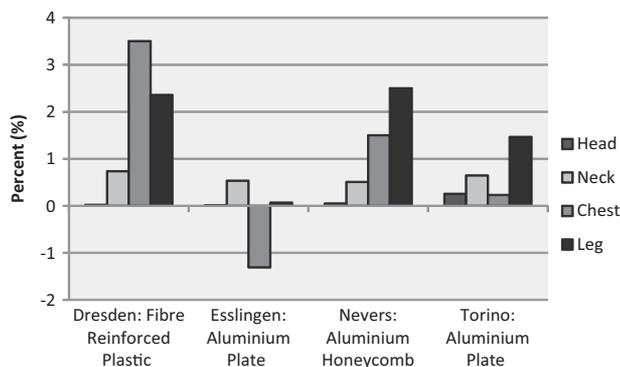
increase in the AIS2 + neck injury risk observed in Fig. 3.

The risk of an AIS2 + chest injury ranged from 17.3 per cent to 35.2 per cent, with the baseline test injury risk at 22.3 per cent (Fig. 3). The change in the chest injury risk due to the crash pulse was between +3.5 per cent and –1.25 per cent and can therefore be considered a contributory factor. The higher value was for the Dresden crash pulse (late peak) and the lower value for the Esslingen crash pulse (early peak). The Nevers crash pulse (two peaks, one of which was early and the other late) also resulted in a slight increase in the chest injury risk. It would therefore be reasonable to conclude that excluding late peak accelerations (and encouraging early peak accelerations) can reduce the chest injury risk during an impact.

The risk of an AIS2 + injury to the femur was lower than for the neck and the chest, ranging from 0.4 per cent to 12.9 per cent (Fig. 3). Altering the crash pulse was observed to increase the injury risk by between 1.5 per cent and 2.5 per cent. The change in the crash pulse was therefore a factor in the increase in the injury risk observed in Fig. 3. The only exception was the Esslingen crash pulse (an early high peak) in which the increase in the injury risk was negligible.

5.2 The occupant stature

The ANOM for the change in the occupant stature is shown in Fig. 7 for an AIS2 + injury risk and in Fig. 8 for an AIS3 + injury risk. The baseline was the 50th-percentile male occupant.

As for the crash pulse, the change in the head injury risk and the change in the neck injury risk were both negligible (less than 1 per cent). The change in the injury risk was significant for both the chest and the femur body regions.
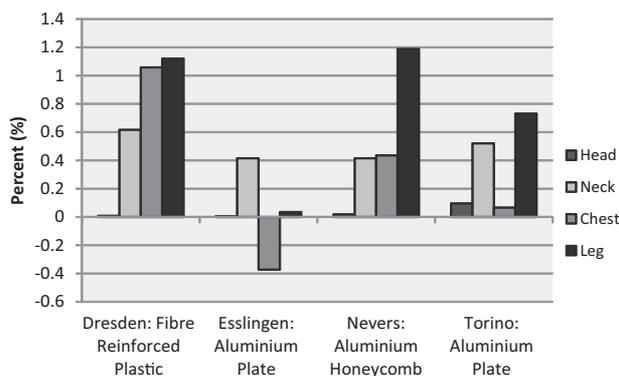

Fig. 5 Change in the AIS2 + injury risk from the baseline (aluminium foam) with change in the attenuator design
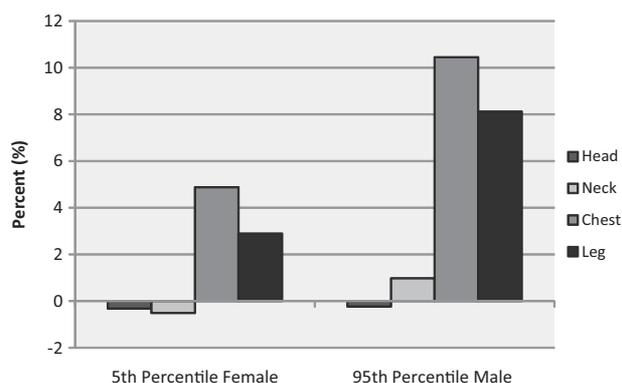

Fig. 6 Change in the AIS3 + injury risk from the baseline (aluminium foam) with change in the attenuator design

Fig. 7 Change in the AIS2 + injury risk from the base-line (50th-percentile male) with change in the occupant size
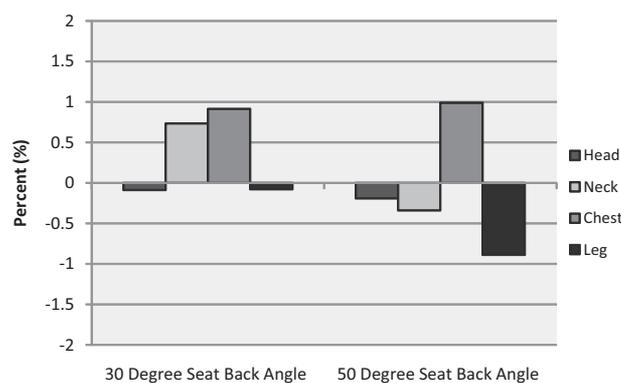

Fig. 9 Change in the AIS2 + injury risk from the base-line (40 ° seat back angle) with change in the occupant posture
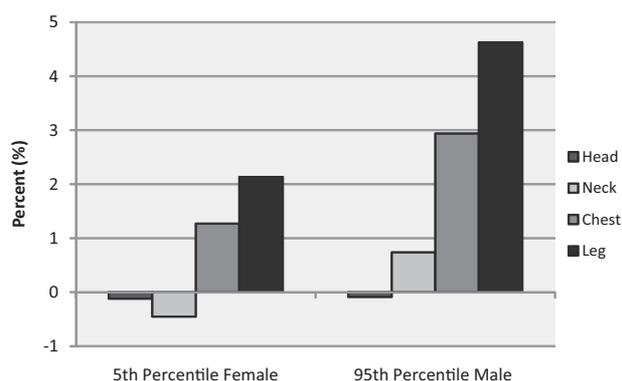

Fig. 8 Change in the AIS3 + injury risk from the base-line (50th-percentile male) with change in the occupant size
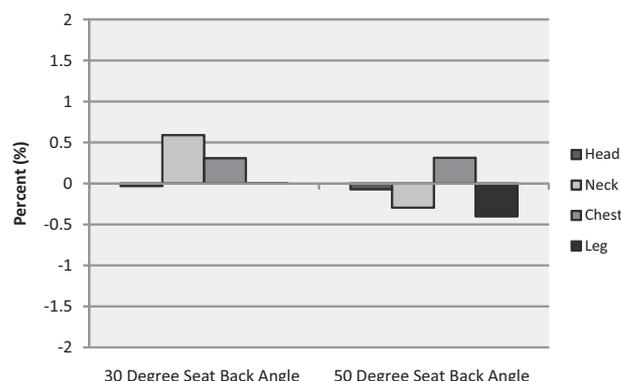

Fig. 10 Change in the AIS3 + injury risk from the baseline (40 ° seat back angle) with change in the occupant posture

For the chest, the injury risk change was + 5 per cent for the 5th-percentile female and + 10.2 per cent for the 95th-percentile male. The occupant stature was therefore a factor in the increase in the AIS2 + injury risk observed in Fig. 3. The reason for this increase is that, in comparison with a passenger car impact in which the seat restraint and supplementary restraints (air bag, knee bolsters, etc.) apply loads to the driver, for this impact scenario the driver was closely coupled to the chassis of the vehicle. In this scenario the seat restraint acting on the driver chest and pelvis is the singular loading device (analysis of the simulations showed that additional contact with the vehicle structure does not occur during the impact). It would therefore be expected that the loads applied by the restraint system to the chest would increase with a heavier occupant and this indeed was the case with average chest loads of 10.8 mm for the 5th-percentile female, 15.4 mm for the 50th-percentile male, and 19.7 mm for the 95th-percentile male. The higher injury risk for the chest

region for the 5th-percentile female was therefore due to the scale factors applied to take account of the different tolerances to injury for occupants of different sizes.

In comparison with a passenger vehicle, the occupant in a single-seater Formula car adopts a straight-leg position (see Fig. 1). As a result, the compression of the femur, which is the cause of the AIS2 + and AIS3 + injuries, was a result of the load applied because the pedals act as a restraint to the forward motion of the leg. This is in comparison with most AIS2 + and AIS3 + injuries in passenger cars that occur because of knee contact with the structure. The average load increased from 1.3 kN for the 5th-percentile female, to 4.6 kN for the 50th-percentile male, and to 5.6 kN for the 95th-percentile male. This clearly indicates that the size of the occupant has a direct influence on the femur load. However, the scale factors applied to take account of the different tolerances to injury for occupants of different sizes show that the risk of injury was slightly higher

Table 5 Neck extension values for the 50th-percentile male (averaged across all crash pulses)

Seat angle (deg)	Neck extension (mm)
30	0.10
40	0.14
50	0.21

for the 5th-percentile female (by approximately 3 per cent) than for the 50th-percentile male

5.3 The occupant posture

The ANOM for the change in the occupant posture is shown in Fig. 9 for an AIS2+ injury risk and in Fig. 10 for an AIS3+ injury risk. The baseline was the 40° seat back angle.

The change in the injury risk for the four body regions was not significant (less than 1.5 per cent). However, an interesting observation was the response of the neck. A change in the posture altered the angle of the head relative to the body and hence the position of the neck, therefore changing the loads experienced by the neck during a frontal impact. The extension of the neck was observed to increase as the angle of the seat was increased (Table 5).

6 CONCLUSIONS

A study has been undertaken that used a computer-aided engineering (CAE) analysis to investigate the injury risk in a Formula Student frontal impact. The computer model was based on the dimensions of the Cardiff Racing CR05 car. The validation of the simulation was based on results obtained from physical tests undertaken by Kettering University. The validation showed that the model was able to predict the injury levels to within 10 per cent of the physical test.

A parametric study was undertaken based on altering the input parameters that could, and would reasonably be expected to, vary between different cars competing at an Formula Student event. These were the crash pulse, the occupant posture, and the occupant stature. While the effects of the crash pulse, occupant position, and occupant stature on the injury risk in frontal impact involving passenger cars have been widely discussed, in motorsport the position of the driver, the addition of a head-supported mass, and the near-direct coupling of the driver's torso to the chassis alters the occupant kinematics during a frontal impact.

The results of the CAE analysis showed that the risk of injury in a frontal impact was dependent on the system variables. The risk of an AIS2+ injury was 22.3 per cent in the baseline constant-g test, increasing to 35.2 per cent in the worst case. This represented a change of +12.9 per cent. The values for an AIS3+ injury risk were 5.1 per cent and 11 per cent, representing a change of +5.9 per cent.

Analysis by body region showed that the head experienced the lowest risk of injury and the least variation in the injury risk. The head injury was negligible in these impacts because of the close coupling of the driver with the chassis. This prevented head contact with solid structures forward of the driver.

The neck was the body region at the highest risk of an AIS3+ injury (up to 11 per cent). This was a consequence of the fact that the torso restrains control of the corresponding crash motion of the driver's helmeted head by holding the shoulders and relying upon the neck actually to restrain the head. This situation was exacerbated by the mass of the crash helmet. An interesting observation was that a change in the occupant posture (the seat angle) altered the extension of the neck (although the change in the injury risk was only slight). This arose because the angle of the head changed the loads experienced by the neck during a frontal impact.

The chest experienced the greatest variation in the injury risk (up to 18 per cent for AIS2+). ANOMs showed that the occupant stature and the crash-pulse shape were responsible for the majority of this variation (up to 10 per cent change and 3.5 per cent change, respectively). In comparison with passenger cars, the seat restraint is the singular method of arresting occupant forward motion in a race car. The increase in the occupant mass caused an additional load to be applied to the chest and hence higher compression and increased injury risk. However, the application of scale factors resulted in a slight increase in the injury risk for the 5th-percentile female compared with the baseline 50th-percentile male. For the crash pulse a late peak was observed to increase the risk of injury compared with an early peak or a constant-acceleration pulse.

The risk of AIS2+ and AIS3+ injuries to the femur was due to the position of the leg which caused the loads applied to tibia by the pedals to be transmitted up through the femur. As for chest injury, an ANOM showed that the femur loading was influenced by the occupant stature (up to 8 per cent change) and the crash-pulse shape (up to 2.5 per cent change). The occupant position in a race car provides for a straight leg and the femur load was caused by the loads applied to tibia by the pedals to be transmitted

up through the femur. This is in comparison with a passenger car in which the femur loads are primarily a result of knee contact with structures in the vehicle occupant compartment.

The improvement in motorsport and passenger car safety has been linking vehicle kinematics to the injury outcome through accident analysis, impact tests, and computer modelling). This has enabled the development of appropriate countermeasures. This study extends this approach into Formula Student and represents a significant step in the process of evaluating and improving Formula Student safety. Cardiff Racing will look to continue this research. The next phase will be to conduct physical tests and further CAE analysis at the higher speeds typical of a Formula Student event.

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APPENDIX 1

Crash pulses

The acceleration–time histories for all the crash pulses are shown in Fig. 11.

Entrant: Essilgen, material: aluminium foam, pulse: constant

Entrant: Nevers, material: aluminium honeycomb, pulse: early and late peak

Entrant: Essilgen, material: aluminium plate, pulse: early peak

Entrant: Dresden, material: fibre-reinforced plastic, pulse: late peak

Entrant: Polytechnic di Torino, material: aluminium plate, pulse: variable
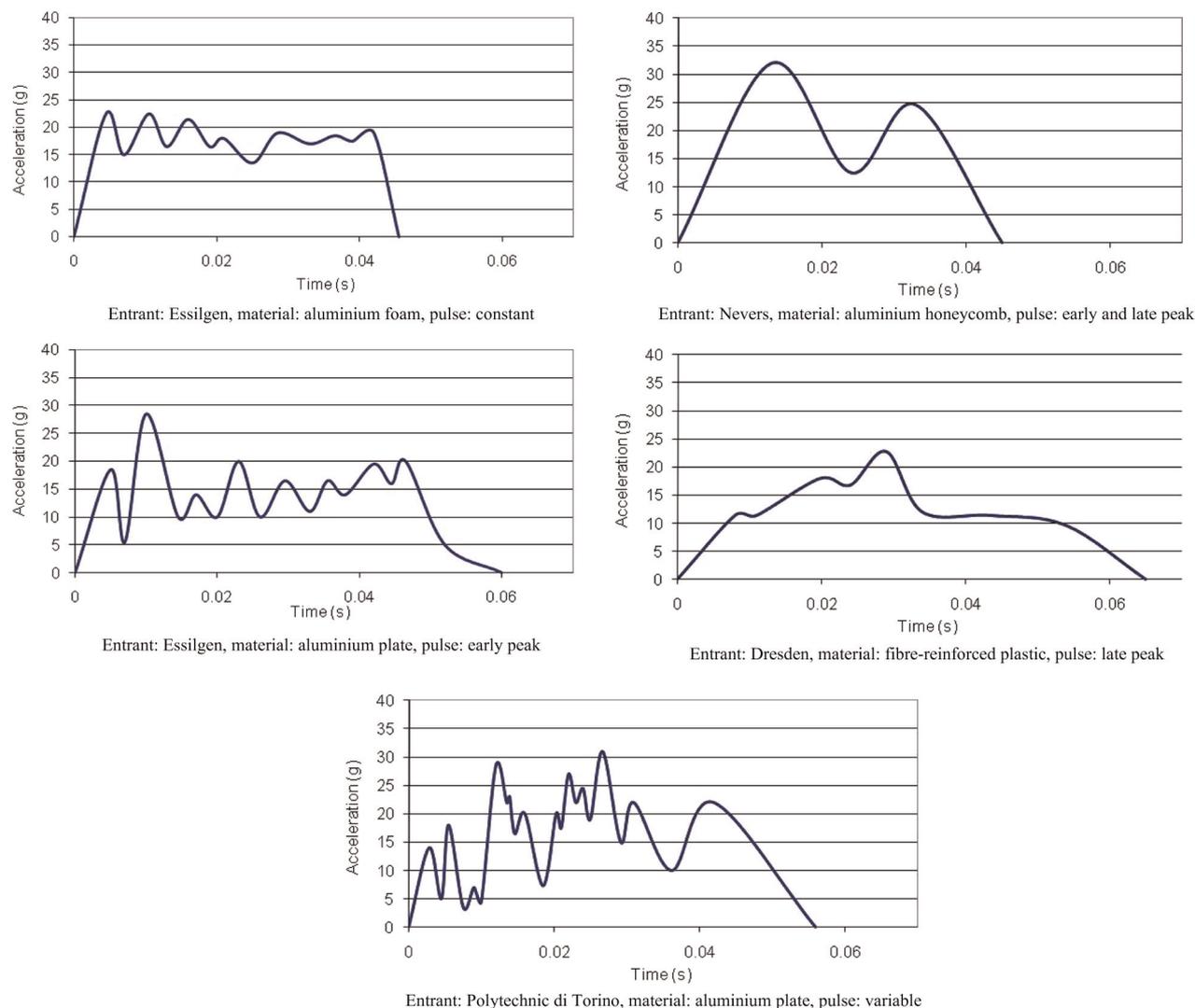
Fig. 11 Acceleration–time histories for the crash pulses

APPENDIX 2

Results set

The full results set is given in Table 6.

Table 6 The full results set

Dummy	Seat angle (deg)	Acceleration pulse	HIC 15 ms	N_{TE}	Chest compression (mm)	Femur compression (kN)	AIS2 + head	AIS2 + neck	AIS2 + chest	AIS2 + upper leg	AIS3 + head	AIS3 + neck	AIS3 + chest	AIS3 + upper leg
5th-percentile female	30	Esslingen 1	50	0.39	8.8	1.37	0.2	16.9	17.5	0.5	0.1	7.8	3.3	0.9
		Esslingen 2	59	0.31	9.1	1.31	0.4	15.7	17.7	0.5	0.1	6.8	3.4	0.9
		Nevers	59	0.50	10.5	1.77	0.4	19.0	18.5	0.6	0.1	9.7	3.6	1.0
		Dresden	59	0.58	11.6	1.69	0.4	20.3	19.1	0.6	0.1	11.0	3.7	1.0
		Torino	74	0.36	9.3	1.42	0.8	16.4	17.8	0.5	0.3	7.4	3.4	1.0
	40	Esslingen 1	52	0.25	11.2	0.99	0.2	14.7	18.9	0.4	0.1	6.1	3.7	0.9
		Esslingen 2	57	0.38	10.1	0.93	0.3	16.8	18.3	0.4	0.1	7.8	3.5	0.8
		Nevers	63	0.33	12.4	1.28	0.5	15.9	19.6	0.5	0.2	7.0	3.8	0.9
		Dresden	61	0.25	13.7	1.29	0.4	14.8	20.3	0.5	0.2	6.1	4.0	0.9
		Torino	83	0.25	11.1	1.37	1.1	14.8	18.8	0.5	0.4	6.1	3.6	0.9
	50	Esslingen 1	40	0.20	9.5	1.03	0.1	14.0	17.9	0.4	0.0	5.6	3.4	0.9
		Esslingen 2	43	0.17	8.3	0.98	0.1	13.6	17.3	0.4	0.0	5.3	3.3	0.9
		Nevers	44	0.24	10.7	1.14	0.1	14.6	18.6	0.5	0.0	6.0	3.6	0.9
		Dresden	49	0.25	11.8	1.14	0.2	14.8	19.2	0.5	0.1	6.1	3.7	0.9
		Torino	57	0.18	9.7	1.47	0.3	13.7	18.0	0.5	0.1	5.3	3.4	1.0
50th-percentile male	30	Esslingen 1	27	0.24	13.8	4.25	0.0	14.6	22.1	2.7	0.0	6.0	4.5	2.7
		Esslingen 2	43	0.31	13.2	4.02	0.1	15.7	21.7	2.4	0.0	6.8	4.4	2.5
		Nevers	30	0.24	15.6	5.47	0.0	14.6	23.5	5.0	0.0	6.0	4.9	3.9
		Dresden	33	0.29	18.2	5.74	0.0	15.4	25.7	5.7	0.0	6.6	5.5	4.3
		Torino	50	0.33	14.6	4.95	0.2	16.0	22.7	3.8	0.1	7.1	4.7	3.3
	40	Esslingen 1	20	0.16	14.0	3.96	0.0	13.4	22.3	2.3	0.0	5.1	4.5	2.4
		Esslingen 2	36	0.24	11.4	4.11	0.0	14.5	20.3	2.5	0.0	5.9	4.0	2.6
		Nevers	21	0.20	15.9	5.31	0.0	14.0	23.8	4.6	0.0	5.5	4.9	3.7
		Dresden	20	0.25	18.3	4.86	0.0	14.7	25.8	3.7	0.0	6.1	5.5	3.2
		Torino	35	0.22	13.9	4.86	0.0	14.3	22.2	3.7	0.0	5.8	4.5	3.2
	50	Esslingen 1	27	0.34	15.2	3.59	0.0	16.2	23.2	1.9	0.0	7.2	4.8	2.2
		Esslingen 2	39	0.39	15.1	3.72	0.1	16.9	23.1	2.1	0.0	7.8	4.8	2.3
		Nevers	25	0.32	16.8	4.97	0.0	15.8	24.5	3.9	0.0	6.9	5.2	3.4
		Dresden	23	0.27	19.2	4.81	0.0	15.0	26.5	3.6	0.0	6.3	5.7	3.2
		Torino	46	0.40	15.3	4.50	0.1	17.1	23.3	3.1	0.1	8.0	4.8	2.9
95th-percentile male	30	Esslingen 1	37.589	0.32	20.0	5.04	0.0	15.8	29.2	6.1	0.0	6.9	6.6	4.5
		Esslingen 2	34.469	0.38	17.1	4.91	0.0	16.8	26.3	5.7	0.0	7.8	5.7	4.3
		Nevers	31.818	0.33	22.4	6.18	0.0	16.0	31.7	11.6	0.0	7.0	7.4	6.8
		Dresden	23.479	0.39	25.5	6.27	0.0	17.0	35.2	12.1	0.0	7.9	8.6	7.0
		Torino	48.847	0.40	20.4	5.61	0.2	17.2	29.6	8.5	0.1	8.0	6.7	5.5
	40	Esslingen 1	59.899	0.43	16.6	5.24	0.4	17.7	25.8	6.9	0.1	8.5	5.5	4.8
		Esslingen 2	38.695	0.40	13.7	5.24	0.1	17.1	23.2	6.9	0.0	8.0	4.8	4.8
		Nevers	59.403	0.44	19.0	6.40	0.4	17.8	28.2	12.9	0.1	8.6	6.2	7.3
		Dresden	46.574	0.42	21.8	6.22	0.1	17.4	31.1	11.8	0.1	8.3	7.2	6.9
		Torino	61.339	0.47	16.9	5.87	0.4	18.3	26.1	9.7	0.2	9.1	5.6	6.1
	50	Esslingen 1	33.169	0.28	19.3	4.56	0.0	15.3	28.5	4.6	0.0	6.5	6.3	3.8
		Esslingen 2	26.017	0.35	16.5	4.92	0.0	16.3	25.8	5.7	0.0	7.3	5.5	4.3
		Nevers	36.229	0.30	21.4	5.73	0.0	15.5	30.6	9.0	0.0	6.7	7.0	5.8
		Dresden	32.471	0.32	24.6	5.71	0.0	15.8	34.1	8.9	0.0	6.9	8.2	5.7
		Torino	41.316	0.37	19.8	5.70	0.1	16.6	29.0	8.8	0.0	7.6	6.5	5.7