The Design of a Structural Hyper-Resisting Element for Life-Threatening Earthquake Risk (SHELTER) for Building Collapse Scenarios: The Safety Chairs

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Abstract: Project SHELTER, Structural Hyper-resisting Element for Life-Threatening Earthquake Risk, aims at developing a strong and stiff functional unit to protect its occupants in case of severe earthquakes that lead to structural collapse. In case of collapse, these units will suffer impacts, particularly if they are installed in upper floors. To avoid severe injuries or death of occupants caused by collapse, safety chairs were designed, provided with shock-absorber systems and auxiliary retaining devices, to keep the occupants properly seated and safe. Three downfall scenarios were evaluated, consisting of vertical and tilted positions. A comprehensive numerical model to represent the human body was developed, mainly focused on chest behaviour and considering the anatomic limits of the vertebral spine. The mechanical ability of the safety chair to ensure the occupants’ safety was evaluated under these harsh conditions. Experimental downfall-and-impact tests will later be performed on shelter units, with crash-test dummies seated on the safety chairs for final validation.

Keywords: seismic shelter; earthquakes; life protection; building collapse

1. Introduction

Fear of earthquakes is usually associated with the risk of building collapses. When buildings are not designed to prevent those collapses, their occupants may die or suffer severe injuries when an earthquake occurs.

Seismic retrofitting of vulnerable buildings is the most appropriate solution, but its costs do not allow for intervention to most of them. Once the main concern in the seismic codes was people’s safety, a proposal for a seismic shelter was presented in [1], which works as a “cocoon” to protect the occupants of seismically vulnerable buildings in case of earthquake occurrence.

Other concepts for seismic protection systems have been proposed without any further development, neither numerical nor experimental [2–4]. The shelter structure was designed to resist, without significant deformation, the effects of the impacts to which it is subjected [5]. There, the life-saving capsule was designed to ensure higher chances of survival for its users, and soon it was understood that specific equipment to protect the human bodies from the shock effects was needed. In this sense, safety chairs were designed to be connected to the main shelter structure by a system of springs and dampers and equipped with anatomical restraint devices. It was also realized that it is difficult to restrain ourselves to a safety chair in the reduced amount of time available, even if an Earthquake Early Warning System (EEWS) is installed with the life capsule, as presented in [1]. This
means that the chair and the human body interact and have differentiated behaviours to respond to the collapse shock accelerations and corresponding inertial forces on the human body. The life capsule structure responds with strength and stiffness to the collapse shock loads, ensuring its integrity. A specific study that assumes this scenario was developed and is now presented in this paper.

In case of building collapse, the shelter and the building debris will interact with each other in a series of shocks until the final collision on the ground. The design of the safety chairs was quite complex and challenging, involving diverse scientific fields such as biomechanics, ergonomics and engineering. The safety chairs had to ensure the following main conditions: being able to oscillate in turn of its initial position, in order to reduce the accelerations of the human body to acceptable levels; being able to withstand the deformations associated with that oscillation; being relatively light; allowing for an easy and immediate occupation upon arrival of shelter occupants; ensuring that retention devices are immediately operational; and having low functional and aesthetical impact, given its preferential location in the entrance hall or corridors of the dwellings [1,5].

Besides the mechanical effects of fall and impacts, the safety chairs were conceived considering different aspects (that will subsequently be validated in experimental testing), such as the psychological reaction to the earthquake alarm, the need for a prompt settling, the adequacy and redundancy of the restraint devices, and the ease to withdraw from the chair after the building collapses.

The present paper focus on the analytical study carried out to design and analyse the mechanical behaviour of the safety chairs.

2. Presentation of the Safety Chair

The more adequate occupants’ position to protect them from shelter impact phenomena during building collapse was studied [6]. Since the time available to get into the shelter is short [1] and panic situations are highly probable, sitting in the chair should be as quick and simple as possible, which led to the adoption of the horse-back riding posture. In addition, to keep occupants away from the periphery of the shelter, where they are more likely to shock with the structure or with building debris, the safety chairs were positioned in a central location inside the shelter. Considering all these aspects, a safety chair structure was designed, constituted by a triangular frame with pinned connections with sub-horizontal elements where the occupant sits (“strut”, in Figure 1), a sub-vertical element where their chest rests (“bridge”, in Figure 1), and a vertical element (“column”, in Figure 1), completing the triangular layout.

Figure 1. Chair structure and body posture.
The chair frame elements are square hollow section steel profiles with 40 mm sides and a thickness of 5 mm ("columns") or 4 mm ("bridge" and "strut"). A high steel grade was considered, with a yield stress of about 1.250 MPa to minimize the weight of the chair.

The columns of the safety chairs are connected to the shelter structure through a shock-absorber system aimed at softening the occupant’s accelerations due to the impacts that may occur during the building collapse, protecting them from severe injuries or death. This system must remain functional during the whole building collapse process, for the various collisions that may occur. This solution is diverse from that used in cars, in which parts of its chassis are used as shock-absorbers to slow down the cabin [7]. In this case, the structure of the chair was designed to remain in the elastic regime, while the reduction of the impact effects is guaranteed by the shock-absorber system composed of a viscous damper and a set of rubber springs (Figure 2).

The viscous damper is installed between the chair’s structure (bottom connection) and the shelter structure (top connection) with pinned connections at both ends to avoid bending forces. A commercial solution was adopted, with a maximum stroke of 600 mm, axial force capacity of 50 kN and an adjustable damping coefficient. The top connection to the shelter structure has an additional steel U profile (UPN 100), welded to the shelter columns. This and other U-shaped profile have two bumpers (Figure 2, detail A), with a global stiffness of 1 kN/mm and capability to shrink 40 mm, aimed at preventing the chair to move outside the shelter limits.

To complete the system, a spring set of three pairs of neoprene blocks are installed (Figure 2, detail B), one pair at the top, one intermediate and one at the lower connection point. The springs make use of the high distortional capacity of neoprene, as seen in bridge bearing pads or in earthquake-based isolation systems. The neoprene blocks present a shear modulus (\(G_{\text{neoprene}}\)) of 0.3 MPa and a damping coefficient (\(\zeta_{\text{neoprene}}\)) of 10%. The blocks at the middle and bottom positions are cylinders with Ø 60 × 100 mm length, while the neoprene blocks of the upper position present a square section of 67 × 67 mm\(^2\) and a length of 160 mm. Aiming at reducing the movement of the upper fixation point of the chair in the horizontal direction to an acceptable level, the effective (free) length of the top pair of springs was reduced by 50%.

Besides the shock-absorber system, additional retention devices will have an important role in complementing the occupants’ protection. The retention system will include cushion devices that not only reduce the body’s acceleration but also avoid the direct contact with the steel elements of the chair structure, conveniently accommodating the human body before any impact. Airbag-type devices were not a choice, given the fact that successive impacts may occur. The cushion system includes a seat cushion, a pillow for the occupant’s chest (chest pillow), a pair of pillows for the occupant’s back (backs pillows) and two pairs of side pillows (side pillows), as shown in Figure 3. The back and the sides cushions are
fixed to two back supports (back braces), in solid Ø 24 mm tubular profile, being a part of the chair structure. These supports, which are retracted (open) while the safety chair is not in use to avoid obstructing the space inside the shelter, may be quickly closed by the occupant after sitting in the chair.

![Figure 3. Chair restraint devices (pillows).](image)

The material of these cushions will behave similar to a crushable foam, capable of absorbing energy, without recovering deformations at least immediately (i.e., it has a plastic behaviour). Soft foams will be used for the seat while hard foams will be used for all the rest. The mechanical characteristics of the materials are specified in Table 1 and Figure 4 (characteristic curve of polyurethane foams or honeycomb-type pillows [1]). It was admitted that the elastic part of these materials’ response is linear and isotropic, while the plastic part follows the behaviour of the foam model developed by Deshpande and Fleck [8]. With the presented characteristics, a numerical model of the safety chair was developed. The designed safety chair weighs a total of 640 N, including the shock-absorber system.

### Table 1. Material conditions for the crushable foams.

| Foam Pillow | Young Modulus E (MPa) | Elastic Poisson Ratio $\nu_{el}$ (−) | Plastic Poisson Ratio $\nu_{pl}$ (−) |
|-------------|-------------------------|-------------------------------------|-------------------------------------|
| Seat        | 0.50                    | 0.30                                | 0.00                                |
| Chest       | 20.0                    | 0.30                                | 0.00                                |
| Back        | 80.0                    | 0.30                                | 0.00                                |

![Figure 4. Pillows characteristic curve.](image)
3. Human body Representation

The design of the safety chair involves the representation of the human body. Based on the well-known GEBOD program [9] and on another study [10], the human body model was built, in terms of geometry and mass distribution, comprising the following 16 units: The head (1); neck (1); chest with abdomen (1); forearms (2); arms (2); hands (2); hip/pelvis (1); thighs (2); legs (2); feet (2).

The representation of the human body is projected on its sagittal plane [9,10], as well as its movement characteristics. The units corresponding to the upper and lower limbs were moved perpendicularly to this plane, according to the representation of the GEBOD program. In general, the body units were modelled as rigid bodies, with a mass on its centre and one or two joints that allow for a connection with the other body units. Only the chest had a more complex representation, in order to reasonably represent the spine behaviour.

Figure 5 presents the assembly of the human body model projected in the sagittal plane, with reference to the centre of mass (CM) and the number of connection joints (J1 or J2) of each element. The geometric characteristics and mass values are presented in Table 2. In Figure 5, the vertical and horizontal axes (YY and XX, respectively) refer to the base of the outer face of a vertical wall, while the human body leans against it in the reference anatomical position (straight-up position). The model has a total mass of 78 kg and is 1.78 m tall in this position, which corresponds to the 50th percentile of a male according to the GEBOD program [9]. As an average male model is heavier than female and infant models, it is expected that the male will suffer higher inertial forces and more severe injuries. Further studies, however, should also comprise those other situations.

According to the consulted bibliography, the behaviour of some joints presented in Figure 5 is well known, namely the head–neck joint (known as atlas and axis joints [11]), the chest–forearms (glenohumeral joints [12]), the chest–hip (sacroiliac joint [13]) and the hip–thighs (femoral capsules). Except for the chest, the assembly of the body units was established through the use of two types of connections: an articulated one that allows for free rotation between the connected nodes, and a rigid one that does not allow for such rotation (Figure 6a). Between the centres of mass of the forearms and arms, an elbow restraint was established to avoid an angle higher than 180 degrees between those parts.
Table 2. Human body features.

| Body Part      | Mass  | XX Coordinate | YY Coordinate | Joint 1 (J1) | Joint 2 (J2) |
|----------------|-------|---------------|---------------|--------------|--------------|
|                | Mass  | XX Coordinate | YY Coordinate | XX Coordinate | YY Coordinate |
| Head           | 4.1 kg| 133 mm        | 1626 mm       | 147 mm       | 1638 mm       | NA           |
| Neck           | 1 kg  | 129 mm        | 1566 mm       | 147 mm       | 1638 mm       | 124 mm       | 1544 mm       |
| Thorax         | 23.8 kg| 133 mm       | 1321 mm       | 124 mm       | 1544 mm       | 146 mm       | 1044 mm       |
| Abdomen        | 2.3 kg| 114 mm        | 1087 mm       |               |              |              |
| Forearms       | 11.2 kg| 100 mm       | 1279 mm       | 100 mm       | 1411 mm       | 100 mm       | 1147 mm       |
| Arms           | 2.8 kg| 100 mm        | 1011 mm       | 100 mm       | 1147 mm       | 100 mm       | 875 mm        |
| Hands          | 1 kg  | 100 mm        | 834 mm        | 100 mm       | 875 mm        | NA           |
| Hip/Pelvis     | 3.6 kg| 151 mm        | 1006 mm       | 146 mm       | 1044 mm       | 164 mm       | 911 mm        |
| Thighs         | 18.8 kg| 160 mm       | 709 mm        | 164 mm       | 911 mm        | 154 mm       | 464 mm        |
| Legs           | 7.6 kg| 154 mm        | 340 mm        | 154 mm       | 464 mm        | 154 mm       | 110 mm        |
| Feet           | 1.8 kg| 161 mm        | 71 mm         | 154 mm       | 110 mm        | NA           |

Figure 6. Modelled human body joints.

In the hands, head and neck, rigid joints were used to prevent free body rotations of these units. The head and neck will additionally be studied in experimental tests, as these parts are particularly sensitive in terms of human safety [12].

The human body model was aimed at estimating the forces generated in the column due to the accelerations caused by the impacts during building collapse. Each body unit undergoes different acceleration values; therefore, there are relative accelerations between them (Figure 6a), which result in internal inertia forces of the human body. In this situation, the spine has a primary role in the transfer of these forces between the lower and upper parts of the body and, in case of extreme stress, this can induce severe and permanent damage to it [13–16].

To reasonably model the influence of the spine on the overall movement of the human body, the chest was modelled with 18 rigid parts (Figure 6b), the bottom five (5) with the same heights of the lumbar vertebrae of the spine, the next twelve (12) with the heights of the thoracic vertebrae and one last to complete the body unit (resembling the cervical part of the human spine). The masses of the chest and abdomen were distributed into a set of points to maintain the position of their centres of mass (Table 2), but at the same time...
distributing this mass along the plane of the body unit and thus endowing it with credible rotational inertia.

The deformation of the chest is mainly, but not exclusively, controlled by the spine. Despite an enormous amount of information available, there is still no consolidated knowledge on the interaction between the various “devices” that constitute or interact with this body unit (spine, rib cage, abdomen, iliac, among others). The developed human body model made use of this disperse information, combining it in a credible way.

To obtain a simplified and reliable computational model of the human body, kinematic restrictions were imposed between the eighteen (18) parts that constitute the chest. This quantifies the forces generated during human body acceleration, by directly associating the movement and deformation of the chest elements with the corresponding inertial forces (Figure 6b). This approach allows for a quantitative analysis of the load transfer between the lower and upper parts of the human body, which is important information to assess the risk of spine damage, as intended.

The overall movement of the chest in the sagittal plane comprised a set of relative movements between its constituent parts (Figure 7), with a kinematic constraint associated with each of them:

- Movement between the spine vertebrae along its axial direction;
- Relative rotations between vertebrae;
- Deferred movements between vertebrae by rotation between them;
- Movements in the sternum of the rib cage, discretized in the corresponding mass points on the periphery of the chest (Figure 6b), that is, only in the frontal ones;
- Movements in the abdomen, discretized at the corresponding mass points on the periphery of the chest;
- Global distortion of the chest.

To compute the displacements and rotations between the vertebrae [17–20], it was considered that the intervertebral discs concentrate all the deformation capacity of the spine. An axial stiffness $k_{2-2}$ of 1000 N/mm was assumed (Figure 7b). Similarly, a rotational stiffness $k_{4-4}$ of 140,000 N-mm/rad was established for rotations between thoracic vertebrae (Figure 7c) and of 35 N-mm/rad when a lumbar vertebra is involved in the rotation (Figure 7d). There is also a cross stiffness ($k_{2-4}$) present, defined in the model by an axial stiffness, with an offset of 40 mm [20] from the column axis (Figure 7e). The linear stiffness has a value of 25 N/mm, resulting in a $k_{2-4} = -1000$ N-mm/mm. The term $k_{2-4}$ is negative as a result of the assumed positive directions on the vertebrae movement (Figure 7a).

The axial direction (stiffness $k_{2-2}$) is the only one with non-linear behaviour. Under compression, the behaviour is elastic up to a 7 mm deformation, associated with a 7 kN compressive force. This force was considered as the ultimate axial compressive load that a vertebra can bear [21–25], in accordance with the preliminary design of the shelter structure [5] and with the limit used by entities such as the USAF (United States Air Force [26]) or FAA (Federal Aviation Administration [27]).

When the chest is stretched in the axial direction of the spine, the muscle tissue, the adipose tissue, and the skin itself offer significant resistance to deformation. In the absence of a valid reference for this tension stiffness, a value 20 times higher than the compressive stiffness was assumed. For imposed deformations, an overestimation of the stiffness value leads to a conservative overestimation of the tension force installed in the spine.

The rib cage and the sternum not only play a fundamental role in stabilizing the spine, but also complement the latter in the forces transfer between the upper and lower parts of the human body. This contribution differs depending on the type of loading to which the chest is subjected (pure compression, bending/extension, lateral bending or torsion). In the present study, as it will be mostly compressed and, in this situation, the ribcage contributes with about 20% to the axial stiffness of the chest [28,29], a stiffness of 250 N/mm ($k_{ribcage}$) was assumed in the model. The points where this stiffness is considered (Figure 7f) correspond mainly to compression in the sternum, which will be the closest scenario to the real situation when the body tends to move to the fetal position (brace for impact). For the
abdomen, it is only possible to qualitatively model its behaviour [30–32] and 100 N/mm is an acceptable value for its stiffness ($k_{\text{abdomen}}$), which is about 10% of the stiffness of the spine, with a similar discretization for the rib cage/sternum (Figure 7g).

Figure 7. Kinematic restraints for the chest.
Since the chest is represented by eighteen (18) parts and with the presented kinematic restrictions, it still needs a restriction that unifies the entire movement, by controlling its distortion in the transverse direction to the axis of the column. This restriction, assumed by the intervertebral discs together with muscle tissues, adipose tissues, and the skin, was modelled as a string when stretched. In this situation, the kinematic constraint needs to allow for the flow of material along the string [25], and for the geometry of Figure 7h, this mainly controls the distortion of the chest in the direction transverse to the axis of the column. Assuming a large distortion capacity in this kinematic constraint, a low elasticity of 500 N was adopted, distributed along its length (L_string). In addition, to avoid a constant flow of the material throughout the entire string, some friction was admitted as reasonably low and depending on the angle between its sections.

While the adopted mass distribution controls the inertia of the chest, the kinematic restrictions, as shown, control its stiffness. To complete the dynamic equilibrium of this body unit, it is still necessary to define its viscosity, a characteristic that is often associated with damping. In the present study, the viscosity, which has a low value [33] and depends on the mass distribution, was associated with the presented springs through damping coefficients in the order of 10% of the critical damping (C_c = 2·(k·m)^0.5). The viscous forces arising from the deformation velocity of the chest are expected to be small when compared with the elastic and inertial forces.

To complete the representation of the human body, it is necessary to define its position when sitting in the chair, which obviously depends on the chair-designed shape, as presented in Section 2. This and the remaining definitions necessary to complete the mathematical model used in this study are presented in the next section.

4. Numerical Model and Study Conditions

Different downfall conditions were considered for the design of the safety chairs as already assumed for the design of the shelter structure [5] (Figure 8):

- Situation 1: Effects of a vertical fall on a deformable ground/debris;
- Situation 2: Effects of a tilted shelter fall when the occupant is seated on a “raised” chair (i.e., on the chair more distant to the shelter impact edge) on the same ground/debris;
- Situation 3: Effects of a tilted shelter fall when the occupant is seated on a “lowered” chair (i.e., on the chair closer to the shelter impact edge) on the same ground/debris.

In the analysis, the occupant was modelled as seated on the cushion with the front of his chest leaning on a pillow (chest pillow), with the lower limbs bent to insert the feet into specific supports (feet restraints, Figure 1), while the upper limbs were bent hugging the “bridge” of the chair.

In general, the interaction between the human body and the retention devices (cushions) was simulated with rigid (without penetration) and frictionless contact surfaces.

Because the effect of the bumpers in preventing the movement of the chair to the outside of the shelter has caused ricocheting of the human body back, an additional pillow was considered for the headrest (Figure 8a), with the same features as the chest pads. This backrest will later be developed through an anatomical study for the retention of head movement and neck stability to avoid the so-called whiplash effect [34]. Two additional kinematic restrictions were also required, one for the feet (Figure 8c) to keep them in their clogs and the other for the arms (Figure 8b), which simulated not only hugging of the chair bridge, but also inlaying against the thighs.

For the analysis of the chair, a time series of displacements were imposed in the points where the chair is connected to the columns of the shelter. Those displacements were obtained in the study for the design of the shelter structure [5]. Therefore, the deformable ground characteristics are the same as in [5], namely a Young’s modulus—E_ground—of 55 MPa and a damping factor—ξ_ground—of 30%. In all the falls, an impact vertical (yy) velocity of 13.4 m/s was considered. This velocity was imposed on the human body and on the chair itself, and all analyses were 1 s long after the impact. Two-dimensional sagittal analysis was performed, restricting all movements out of that plane.
Figure 8. Chair positions for tilted falls considered (at left—“lower” chair; at right—“upper” chair).

In the former author’s study [5], the mass of the chairs and their occupants was represented as a unique lumped mass, obtaining a first estimate of the accelerations that they will be subjected to during the impact of the shelter. An acceleration value around 70 g was then obtained, considering a unique lumped mass for the chair and its occupant. However, the movement of the occupant may differ considerably from that of the chair, particularly in respect to accelerations/inertial forces.

Conversely, the limits of occupant survival assumed when designing the structure of the shelter were based mainly on experimental results [35,36] involving the so-called
Hybrid III dummy [37]. This physical model has been shown to be highly reliable for situations involving frontal and lateral impacts of the human body, but it is not that good in impacts involving compressions of the spine [38,39], a situation that sometimes requires the modification of the test dummy.

To limit the maximum occupant’s acceleration, the shock-absorber system considered in [5] needed to be redefined. The characteristics of the damper were maintained, while the stiffness of the rubber springs was reduced to ensure survival conditions, particularly regarding spinal compression.

Finally, although the numerical model focuses on a single anatomical reference (50th percentile of a male in the GEBOD program), the adopted solution of the chair can be easily adapted to other anatomies (Figure 9 shows an adaptation for a child), for example, different gender, age, or weight, etc, namely by adapting the back pillows.

Figure 9. Chair solution and its fitness for child anatomy (1.40 m tall).

5. Study Results and Discussion

The results for the three (3) fall situations considered in the study present similarities between them. The most important concerns the system that dampens the shock effects, with large reductions in the accelerations suffered by the human body, when compared with those of the shelter structure. The damper device is more effective in the vertical direction (YY) and in the initial moments of the shelter impact (when the velocities are higher), while the neoprene springs and the bumpers are more effective in the horizontal direction (XX), even when the falls are tilted. These bumpers are also needed to keep the chair inside the shelter, despite that parts of the chair, and parts of the human body (only the feet), may theoretically pass beyond the shelter external limits due to the distortion of the lower and intermediate neoprene springs.

Figure 10 shows the positions of the chair and the human body at different instants of the shelter vertical fall (situation 1). Since the results for all the considered situations are similar regarding the movement of the chair and body, those images are also representative of the other fall situations.

The movement is, in an initial phase (about half a tenth of a second), essentially vertical, accompanied by clashes between the occupant’s chest and the cushions that surround it. As the damper device reduces the speed of the chair structure and the body adapts to the seat cushion, the damper is much more effective in the vertical direction, by reducing the velocity in this direction, but not in the horizontal one. The chair then shows a tendency to move horizontally outwards the shelter at its lower end, being restrained by the bumpers of the lower UPN100 profile (Figure 2). With these clashes, as mentioned, the body is then projected backwards—the back against its cushions in the first place, followed immediately by the head against its back cushion.
Figure 10 shows the positions of the chair and the human body at different instants of the shelter vertical fall (situation 1). Since the results for all the considered situations are similar regarding the movement of the chair and body, those images are also representative of the other fall situations.

Subsequently, already in the ascending phase of the movement of both the chair and the human body, the latter tends to take off from the seat and to hit the chest cushion, especially in the tilted falls (situations 2 and 3), right after a new impact of the chair columns against the bumpers of the UPN 100 profiles (around 150 milliseconds), with higher severity than the first ones (Figure 11). This severity is mainly due to the almost simultaneous impact of the columns against the top and the bottom bumpers of the UPN 100 profiles, which occurs only for tilted downfalls.

Afterwards, there is a series of bounces against the pillows that surround the human body, repeating and decreasing in intensity as they occur. This upwards movement ends with the impact of the shoulders on the upper part of the back cushions (Figure 10, Time = 0.230 s), followed by a new impact of the head against its back, in a much higher position than in the first impact. Since the initial instant, about four (4) tenths of a second have passed, and velocities as well as accelerations in the chair and in the human body have decreased, reaching much lower values. The vertical downwards movement resumes, not necessarily in the same way, but with the generic features of the first cycle and with less intensity.

The stresses on the chair’s steel elements are always below its yielding limit ($f_y = 1.250 \text{ MPa}$), therefore not exceeding the elastic regime, and ensuring safety during free fall, which is consistent with the situation of several impacts along the fall.
5.1. Human Survival Capacity

The mean accelerations of the human body (weighted by the masses of each body part), through directions XX (horizontal) and YY (vertical), can be compared with human tolerance curves, such as “Eiband curves” [40], plotted based on experimental results, in which astronauts were subjected to high accelerations when seated on a rigid seat and pressed against it by a harness. When using these curves, trapezoidal seat acceleration pulses are compared with the limits established in these curves, where not only the maximum acceleration value of the pulse is important, but also its duration. As in this study, the occupant may suffer some relative movement in relation to the chair, and the comparison pulses were traced such that the area of the diagram with the evolution of accelerations over time coincided with the area of the trapezoidal pulses.

For the situation that leads to higher spine compression (situation 2, “upper” chair, Figure 8), an acceleration pulse with a duration of 63 milliseconds and a maximum vertical acceleration of 22 g is obtained (Figure 12a), reasonably below the limit of human tolerance, which is about 40 g in this situation (limit for “moderate injury”). The acceleration is not constant through the duration of uniform acceleration of the pulse, reaching a sudden peak of 46 g, which lasts less than 2 milliseconds above the Eiband’s human tolerance limit. For the horizontal accelerations, the pulse with the highest acceleration lasts 11 milliseconds and has a maximum acceleration of 43 g (Figure 12b), also below the human tolerance limit, which in this case is about 110 g (“moderate injury”), and it is never surpassed.
5.2. Human Chest Motion Features

Besides the head, it is the chest, and in particular the spine, that is the important body part to protect. The study of the spine’s behaviour during the different situations evaluated in this study can be performed through the analysis of vertical (YY) and horizontal (XX) accelerations. The global axes XX and YY remain fixed throughout the analyses.

For the spine, the upwards vertical accelerations (Figure 13a) are significant at the beginning of each analysis, reaching their highest absolute values when the damper tends to lead the velocity in this direction to zero, which happens before the first tenth of a second. Once these maximums are passed, with the impact on the bumpers, a second set of maximums appears, with an opposite sign and with absolute values greater than the first ones, in the case of tilted falls (situations 2 and 3).

This behaviour arises from the impacts of the body against the chest cushions, which is not the case for the vertical fall (situation 1). Thereafter, the vertical accelerations are reduced, with some episodes caused by the collisions with the chest and back cushions or with the return of the body to the seat of the chair. Directly before the half-second, a new set of peaks of ascending vertical accelerations appears, now corresponding to the second cycle of vertical movement and in the downwards phase, which presents lower absolute values when compared with the first set of peaks.

The horizontal accelerations present high values in the first half (1/2) tenth of a second of the analyses, when the vertical velocities are higher, induced by the collisions of the chest against the cushions that surround it. With the reduction of vertical velocity, these impacts become less intense, and consequently, the values of horizontal accelerations are reduced. For tilted falls (situations 1 and 2), the moment of the impact of the columns of the chair against the bumpers is critical, since the subsequent impact of the chest against its cushions returns the highest absolute values of horizontal acceleration in the spine, which are higher than the peak vertical accelerations. Thereafter, the horizontal accelerations repeat a pattern similar to that of vertical accelerations.

The comparison performed between the average accelerations of the human body, thus considered as a whole concentrated mass, and the limits of the “Eiband curves” are good indicators for human tolerance to the impacts that the body suffers during the building collapse, although the movement of the human body, particularly the chest, has important components of relative translation and rotation between vertebrae (Figure 7) that are not considered in these curves. Thus, these conditions do not allow for accurate assessment of whether the compressions in the spine exceed their tolerated limit, one of the main concerns of this study.

For this reason, the human body model was also used to individually evaluate the axial and shear forces installed in the spine, as well as in the rib cage/sternum and in the abdomen (the latter in absolute value). The tilted fall of the “upper” chair (situation 2) is the situation that returns the greatest compression forces on the spine (Figure 14a), and it is also clear that the lower vertebrae are the most affected (all lumbar, L1 to L5 and some of the thoracic, T6 to T12), which happens for the three analysed situations. It is important to stress that the limit established for the compressive strength of the vertebrae (7 kN) is never exceeded.

The upper thoracic vertebrae (T1 to T5) turn out to be the least compressed, not only because the chest/sternum supports part of the axial load, but also because the rotation of the body to the fetal position (bracing for impact) compresses the thoracic cage more (in particular the sternum) and relieves the compression of the spine. In this movement, the rib cage is subjected to a force lower than 2 kN (Figure 14c). The absolute maximum of the force supported by the rib cage occurs with the frontal impact of the chest against the cushion, when it reaches a value of almost 6 kN.
Figure 13. Spine vertebral accelerations (situation 1 to situation 3).
reduced, with some episodes caused by the collisions with the chest and back cushions or with the return of the body to the seat of the chair. Directly before the half-second, a new set of peaks of ascending vertical accelerations appears, now corresponding to the second cycle of vertical movement and in the downwards phase, which presents lower absolute values when compared with the first set of peaks.

The horizontal accelerations present high values in the first half (1/2) tenth of a second of the analyses, when the vertical velocities are higher, induced by the collisions of the chest against the cushions that surround it. With the reduction of vertical velocity, these impacts become less intense, and consequently, the values of horizontal accelerations are reduced. For tilted falls (situations 1 and 2), the moment of the impact of the columns of the chair against the bumpers is critical, since the subsequent impact of the chest against its cushions returns the highest absolute values of horizontal acceleration in the spine, which are higher than the peak vertical accelerations. Thereafter, the horizontal accelerations repeat a pattern similar to that of vertical accelerations.

The comparison performed between the average accelerations of the human body, thus considered as a whole concentrated mass, and the limits of the “Eiband curves” are good indicators for human tolerance to the impacts that the body suffers during the building collapse, although the movement of the human body, particularly the chest, has important components of relative translation and rotation between vertebrae (Figure 7) that are not considered in these curves. Thus, these conditions do not allow for accurate assessment of whether the compressions in the spine exceed their tolerated limit, one of the main concerns of this study.

For this reason, the human body model was also used to individually evaluate the axial and shear forces installed in the spine, as well as in the rib cage/sternum and in the abdomen (the latter in absolute value). The tilted fall of the “upper” chair (situation 2) is the situation that returns the greatest compression forces on the spine (Figure 14a), and it is also clear that the lower vertebrae are the most affected (all lumbar, L1 to L5 and some of the thoracic, T6 to T12), which happens for the three analysed situations. It is important to stress that the limit established for the compressive strength of the vertebrae (7 kN) is never exceeded.

![Figure 14. Internal forces in the human chest (situation 2).](image)

The axial force in the spine vertebrae reaches its maximum positive value, about 10 kN, in the upper vertebrae (thoracic, T1 to T5), while in the other vertebrae, it does not reach 5 kN. As a reminder, this stretching load is installed not only in the spine but also in the muscle tissues, the adipose tissues, and the skin of the chest.

The values of forces in the abdomen are much lower when compared to those on the spine or on rib cage, slightly exceeding the value of 1 kN when the spine reaches maximum compression (Figure 14d).

As a main conclusion, it may be pointed out that the transverse forces on the spine, in particular on its intervertebral discs, came to be quite reduced (Figure 14b) with the shock-absorber system and the auxiliary retaining devices, as intended.

Experimental tests are currently underway, where the key parameter being measured is the instant peak accelerations of the more important body parts, namely the chest and the head. Several experimental downfall-and-impact tests are being conducted, from nine (9) different heights (one-meter intervals) and in different falling slopes, such as those assumed for the design stage (Figure 8). A non-instrumented Hybrid III crash-test dummy occupies the shelter units, settled on safety chairs. The sensors being used are optical targets captured by two (2) 300 fps HFR cameras, along with twelve (12) 200 g wireless accelerometers [41] to sensor the crash-test dummy, with a sampling rate of 330 Hz.

5.3. Shock-Absorber System

Figure 15 depicts the components of the shock-absorber system, namely the damper device and its endings (highlighted in pink), as well as the neoprene springs and their endings (highlighted in yellow). The rubber bumpers and the UPN100 profiles are also highlighted in red. The system behaved as expected, with the damper device being crucial in the initial moments, when velocity in the chair and in the human body in relation to the shelter is higher (Figure 15b). After those initial moments, when the velocity decreases, the elastomeric springs and the rubber bumpers on the UPN 100 profiles control the chair’s response.
particular the sternum) and relieves the compression of the spine. In this movement, the rib cage is subjected to a force lower than 2 kN (Figure 14c). The absolute maximum of the force supported by the rib cage occurs with the frontal impact of the chest against the cushion, when it reaches a value of almost 6 kN.

The axial force in the spine vertebrae reaches its maximum positive value, about 10 kN, in the upper vertebrae (thoracic, T1 to T5), while in the other vertebrae, it does not reach 5 kN. As a reminder, this stretching load is installed not only in the spine but also in the muscle tissues, the adipose tissues, and the skin of the chest.

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For the damper (Figure 16), its 600 mm maximum stroke proved to be enough, as in the worst situation (situation 1), it deforms about 400 mm downwards (Figure 15a) and less than 150 mm upwards. With a maximum value of 27 kN, the axial force installed in it was well below its 50 kN limit (Figure 16b). The axial and shear forces inflicted by the chair on the shelter columns were also computed (Figure 16c,d).

Figure 15. Details of the shock-absorber system (frontal and lateral views).

Figure 16. Damper results (situation 1 to situation 3).
The elastomeric springs present high distortions (Figure 17), being higher for the upper spring, with a maximum of 600% in the three (3) fall situations. The intermediate and lower springs suffer lower distortions, with maximum values of about 400%.

![Distortions of the superior neoprene springs](image)

![Distortions of the inferior neoprene springs](image)

**Figure 17.** Displacements of the top (superior) and bottom (inferior) neoprene springs (fall situation 1 to situation 3).

The values of the distortional components transverse to the shelter columns differ among each situation, with negative values corresponding to movements outside the shelter limits. For the lower and intermediate springs, this movement out of the shelter depends on the falling situation, being less relevant for the tilted fall on the raised chair (situation 2) and more significant for the vertical fall (situation 1, Figure 10). In the latter, the chair columns theoretically come out of the shelter’s limits for about 110 mm (210–100 mm, where 100 mm corresponds to the gap between those columns and the limits external shelter boundary).

Finally, for the situation where the springs and the damper impose larger bending stresses to the shelter columns (situation 2, Figure 18), this pair of columns will have to be able to withstand an additional bending moment of 51 kN·m, which represents 42% of the pure flexural strength of the corresponding profile (HEB120 in S355 steel grade).
Figure 18. Additional bending moments on the shelter column at \( t = 152 \) milliseconds (situation 2).

6. Conclusions and Future Developments

The design of the safety chair ensures that it is reliable, resisting the forces due to the predicted actions and significantly increasing the chances of survival of its occupants. The solution developed for the chair is relatively light and easy to use for people with different sizes and capacities, while at the same time not interfering with the circulation of the shelter’s corridor when closed (Figure 9).

The model developed for this study was essentially focused on the behaviour of the human body undergoing strong impacts, particularly on its chest and spine, given the risks of severe and permanent injuries. A relatively simple version of a global anatomical model first developed has quantified the internal forces generated with the relative accelerations between different parts of the human body. Three (3) fall situations were analysed, with different inclinations, and similar results were obtained in relation to these. Of all the results presented, in the three (3) situations, the limit considered as bearable for compression in the spine was never exceeded.

The planned experimental simulations of the shelter downfalls will expectably add new information and point out other improvements, namely regarding head and neck protection.

Based on the presented study, several downfall and impact tests are currently underway to assess the safety chair design conditions hereby presented.

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