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COMPARATIVE MULTIBODY DYNAMICS ANALYSIS OF FALLS FROM PLAYGROUND CLIMBING FRAMES

ABSTRACT

This paper shows the utility of multibody dynamics in evaluating changes in injury related parameters of the head and lower limbs of children following falls from playground climbing frames. A particular fall case was used as a starting point to analyze the influence of surface properties, posture of the body at impact, and intermediate collisions against the climbing frame before impacting the ground. Simulations were made using the 6-year-old pedestrian MADYMO rigid body model and scaled head contact characteristics. Energy absorbing surfaces were shown to reduce injury severity parameters by up to 30-80% of those of rigid surfaces, depending on impact posture and surface. Collisions against components of a climbing frame during a fall can increase injury severity of the final impact of the head with the ground by more than 90%. Negligible changes are associated with lower limb injury risks when different surfacing materials are used. Computer reconstructions of actual falls that are intended to quantify the severity of physical injuries rely on accurate knowledge of initial conditions prior to falling, intermediate kinematics of the fall and the orientation of the body when it impacts against the ground. Multibody modelling proved to be a valuable tool to analyze the quality of eye-witness information and analyze the relative injury risk associated with changes in components influencing fall injuries from playground climbing frames. Such simulations can also support forensic investigations by evaluating alternative hypotheses for the sequence of kinematic motion of falls which result in known injuries.

KEYWORDS

impact surface; multibody modelling; monkey bars; injury severity; head injury

1. INTRODUCTION

Playground falls are a public health concern which commonly leads to injury [1-5]. Arm and leg fractures are the most common major injury from falls but the potential consequences of head injury are more severe [6, 7]. Child abuse cases are sometimes falsely reported as accidental falls, therefore biomechanical and forensic investigations of injury suffered by children are often necessary [8, 9].

Previous studies indicate that important risk factors are the height of playground equipment, the posture of a body at impact with the ground, and the type of surface onto which children fall [4, 10-12]. Laboratory studies have been limited to analysing simple free falls [13-16]. Computer simulations, however, allow for tests of more complex scenarios with repeatability and precision, which is difficult with laboratory experiments of free falls using dummies or when reconstructing real world impacts [8, 17-19]. They also make it possible to model impacts involving complex systems such as the human head, which is difficult to do physically and repeatedly [20].

Studies suggest that impact absorbing surfaces in playgrounds reduce the risk of injury [11, 21]. Nevertheless, whenever an injury is sustained in a playground that does not comply with current impact surfacing standards, the question arises as to whether or to what extent an impact absorbing surface would have reduced injury severity.

This question was posed following an injury sustained from a playground fall in which a 6-year-old girl fell from a climbing frame, and hit her head against the ground. According to available medical records, she suffered severe head injury with right parieto-temporal impact and skull fracture, secondary cerebral insults, respiratory distress syndrome and abdominal injuries. It is unclear whether the surface against which she fell was tarmac or concrete. This particular accident is simulated to illustrate several different factors that contribute significantly to head injury risk.

The information concerning the nature of the fall was limited. An eyewitness who was with the victim on the day of the fall stated that “she climbed onto the top of the frame, her foot slipped off one of the bars, and she fell”. He states that “her back hit one of the other metal bars and she fell onto the ground”. The witness simply says that he saw her hit her head (it was not reported whether this was on the side or back of her head) on the surface beneath the frame. Ultimately, the precise sequence of the fall is unclear.

The greatest vertical distance from the highest point of the climbing frame to the impact area below was 2.7m. No further dimensions were provided.

This accidental fall was used in the present analysis because it possesses three important unknowns which illustrate the variability of the possible injury risk associated with factors that should ideally be documented when undertaking forensic reconstructions.

1. *Impact Surface:* The impact on the ground was against a hard surface, but it is unknown whether this was tarmac or concrete.

2. *Fall Posture*: Witness information suggested that the child fell either on her back or side.
3. *Fall Sequence*: Witness information also suggested that the child's body impacted the climbing frame during the fall.

2. MATERIALS AND METHODS

Computer methods allow for repeatability in experiments where the outcome of an experiment needs to be analysed in relation to specific changes in certain input variables, while maintaining other conditions constant. In this study, simulations are conveniently performed using MADYMO [22]: this allows computational impact experiments to identify and quantify the influence of individual variables, in contrast to physical fall reconstructions where it is difficult to control or vary test parameters.

2.1 MADYMO Model

The MADYMO 6-year-old child rigid body pedestrian model was used. This is based on the anthropometry of the 50th percentile 6-year-old child, the height and weight of which are 1.17m and 23kg. The contact characteristics of the model's different body sections can be modified based on any specific experimental data available. The MADYMO pedestrian models have been validated satisfactorily regarding model kinematics, trends and shape of the head output acceleration and forces [23]. The validation simulations and results of the model have been thoroughly described by Hoof et al, [24]. The model is multi-directional, which means that it is applicable for complex impact scenarios. Leg fracture tolerances were also analyzed in detail by Hoof et al., [24]. Illustrative fall reconstructions of the present case study are given on the AVI files that accompany this paper (see graphics files "Back45_impact_position.AVI" and "Front_impact_with_bar_position.AVI").

2.2 Head Contact Characteristics

O'Riordain et al., [25] used values from [26] to model the head contact force-displacement characteristics. Similarly in this study, the values used for the head contact characteristics are based on experiments [26]. Because those experiments were performed on adult skulls, it was necessary to scale the stiffness values to obtain biofidelic values more appropriate to a 6-year-old child (Figure 1). The scaling factor was obtained using the same methods used by the MADYMO's SCALER module [22].

Sensitivity analyses were made using different contact characteristics to analyze their influence on the results. Simulations were run with a more compliant and a more rigid head contact characteristic. As the head contact characteristic was changed, the absolute magnitudes of contact forces and accelerations changed in proportion to the contact characteristic. Nevertheless, changes in the head output values were similar for all tested head contact characteristics when other variables were changed, i.e., impact posture and ground contact characteristics. This means that the methods used in this study, as in the model used, are appropriate to be used for relative comparisons, but not to generate conclusions from isolated output values. The model has been validated in head output trends and body kinematics, but not for absolute values of head impact output. The scaled model based on the characteristics from [26] was chosen as the baseline

model used in this study, as this was considered to be the best approximation for the contact characteristics of a child.

INSERT FIGURE 1 HERE

2.3 Impact Surfaces

The injury risks associated with falling on different stiffness surfaces were studied. For this, it was necessary to use contact stiffness properties for different kinds of surfaces, namely concrete [27], tarmac [28] and rubber [29] (Table 1). From these properties, elastic force-displacement characteristics were obtained via Boussinesq's [30] classical theory of stress in homogeneous semi-infinite masses. For turf stiffness properties, force deflection curves were derived from impact acceleration measurements made on soft turf covered soil using a tri-axial accelerometer attached to a cast aluminium hemisphere [31]. Since turf is not an adequate playground surface [32], its stiffness characteristics have only been used to provide comparison against a considerably more compliant impact surface.

INSERT TABLE 1 HERE

2.4 Impact Orientation

Impact orientation identifies the posture of the body relative to the ground at the instance of first contact during an impact event. The climbing frame had a maximum free fall height of 2.7m. This was used as the worst case scenario since head injury risk is proportional to free fall height [15] and, in turn, to potential energy. This same fall height was used for all simulations, although the initial orientation of the body (e.g. standing vertically, kneeling or lying horizontally) was varied. This means that for a standing position the head would have more potential energy than for a horizontal position, but the body would absorb most of the impact, whereas for a prone position the head would impact the ground at the same time as the rest of the body.

Simulations were run using impact orientation as the only variable parameter. The child model was simulated to fall onto the ground in seven alternative impact postures (Figure 2). Additional simulations including angled or vertical head-first impacts were not considered as these would obviously lead to more extreme cases of head injury.

INSERT FIGURE 2 HERE

Since mid-air collisions can change the impact conditions, it is necessary to analyze mid-air impacts against the climbing frame to see how they influence impact against the ground and consequently head injury related parameters. The child model was allowed to fall from the same 2.7m height from a lying position, but with the difference that before striking the ground, the child's upper back struck a rigid bar 1.5m from the ground. The final impact posture was the same as in Figure 2c, but the kinematics of the fall prior to striking the surface was different.

The variables that were considered were linear and angular head accelerations, HIC score (which is calculated as an integral of the linear acceleration pulse and is commonly used to indicate the likelihood of a person suffering a head injury [33]), head-ground contact force and leg forces (axial and lateral), where relevant. The leg force values were compared between different impact

surface contact characteristics to observe relative lower injury risk associated with a fall from or to a feet-first posture. Since the MADYMO child model has not been validated to the same extent as the adult models, it is appropriate that the predicted results should not be interpreted in absolute terms to assess actual injury risk. Nevertheless, MADYMO is useful to examine the relative change of injury related parameters. In this study changes in the simulation results are examined when initial conditions such as impact postures and impact surfaces are changed. Consequently, the increase or decrease of injury related parameters are shown as percentages.

3. RESULTS

HIC decreases substantially when the surface changes from concrete to rubber or to turf (35% decrease for rubber and 82% decrease for turf for the “front” impact posture (Figure 2(c)). The influence of the impact posture of the model is also clearly observed. When the model faces downwards (Figures 2(c&d)), the HIC values are considerably larger than other impact postures (Figure 3). The posture that gives the highest HIC values is the “front” impact posture (Figure 2(c)). The other postures, (“st45”, “side”, “front45”, “back”, “back 45”; Figures 2(a,b,d,e,f) respectively), give values 69%, 47%, 12%, 41% and 89% lower in the case of the concrete surface. Other surfaces reflect similar changes. Linear acceleration values (not shown) also vary in an equivalent fashion.

INSERT FIGURE 3 HERE

A noticeable decrease (52% decrease for the “front” posture (Figure 2(c)) is observed in angular acceleration values when the surface properties are changed from concrete to rubber (Figure 4). A clear trend is observed; angular acceleration decreases as more compliant ground properties are used. Angular acceleration also varies considerably with impact posture (Figure 4). This is particularly seen in the “front45” posture (Figure 2(d)), in which the angular acceleration for most impact surfaces is almost twice that of the next most severe impact posture, in this case “front” (Figure 2(c)).

INSERT FIGURE 4 HERE

The contact force between the head and ground decreases considerably by changing surface properties from stiff to compliant (37% reduction from concrete to rubber and 73% reduction from concrete to turf for the “front” posture (Figure 5). However, a very small change is observed between concrete, tarmac and the rigid surface (9% and 4% decrease from rigid to tarmac and concrete, respectively, for the front posture (Figure 5)). This is also true for the other injury parameters. The orientation of the model when it strikes the ground also has an appreciable influence on the contact force, but this is less than for the other calculated values. Differences in contact force for different impact postures are smaller than differences in other injury related values. The largest decrease in contact force below that of the concrete surface is 50% while for HIC, linear acceleration and angular acceleration it is 89%, 59%, and 72% respectively.

INSERT FIGURE 5 HERE

Impact against a bar prior to impacting the ground can substantially increase head injury related values. The increase of HIC was 37% for the “front” impact posture and 96% for the “side” impact posture (Table 2). The conditions “Front-with-bar” and “Front-direct-impact” have the same body orientation at impact (c.f. “Front”, Figure 2(c)). In “Front-with-bar”, the body strikes the bar which causes the body to rotate to the same impact orientation as in the “Front-direct-impact” condition, where the body impacts the ground without any mid-air collisions. This applies similarly for the “Side-with-bar” and “Side-without-bar” conditions.

INSERT TABLE 2 HERE

Most head injury parameters increase when the upper back of the model strikes a bar before finally impacting the ground (Table 2). In some cases, they increase by a greater proportion than does the head speed. This is due to the rotation of the body from the impact against the bar. Due to the induced forward or sideways rotation of the body combined with the downward velocity, the final impact with the ground is more severe. The rotation induced by the impact with the bar accelerates the head vertically by a greater amount than when the body falls freely (c.f. 9m/s Vs 7m/s; Figure 6).

INSERT FIGURE 6 HERE

The contact force value in the “Front-with-bar” impact condition in Table 2 is lower than in the other three cases, which is not consistent with the increase of the other head injury related parameters. This is because in this situation the ellipsoid corresponding to the face impacts the ground before the ellipsoid corresponding to the head, and since the face has a much more compliant force-displacement characteristic, the force level is lower. In the “Front-direct-impact” condition, the forehead impacts the ground first, and this corresponds to the head contact characteristic, which is much stiffer.

As expected, the axial leg forces (upper and lower leg forces exhibit similar trends) are higher for the standing impact posture of Figure 2(g) (Figure 7). The leg forces for impacts against the rigid surfaces are very similar, and only a small decrease is seen when the surface characteristics are more compliant, such as for rubber or turf. When the impact surface is changed from a stiff surface, the change in leg forces is much smaller than the change in HIC. When rubber surfaces are considered, HIC decreases by 34-63% (Figure 3) while leg forces in the standing position only decrease by 16-22% (Figure 7). For more compliant surfaces such as turf, these reductions are more pronounced. These trends are also observed for values of upper and lower leg lateral force and for bending moments (not shown) associated with lateral flexion of the leg.

INSERT FIGURE 7 HERE

4. DISCUSSION

4.1 Surfaces

It has been seen that the type of surface material can have a noticeable influence on head injury related parameters. Doubts regarding the reduction of impact severity when using more

compliant surfaces are easily resolved by multibody dynamics, as well as with physical experiments. The value of multibody dynamics simulations is clearly seen since these can be done quickly, and with proper judgement, conclusions can be made regarding a specific fall scenario. All parameters are reduced when rubber or turf are used. This supports earlier studies [4, 34] documenting that energy absorbing surfacing is more important than actual fall height in determining injury risk. Even though turf gives the lowest injury related values, turf cannot be recommended for playground surfaces [32] since its energy absorbing properties decline rapidly with use and repeated impacts. Rubber is a better playground surface material, since it reduces the levels of injury parameters considerably and maintains its impact response characteristics over prolonged traffic from playground users.

HIC scores are found to be much higher for stiff surfaces than for more compliant surfaces such as rubber or turf. The noticeably less rigid force-displacement characteristics of rubber and turf partially explain this phenomenon. The fact that HIC is associated with sagittal plane impacts is also relevant. With this in mind, the HIC values obtained from impacts other than frontal impacts might not be reliable for establishing a possible injury risk associated with a value exceeding 1000, as in the current standards for playground safety [35]. Because of this, a sole value for either HIC or another head injury parameter is not considered. For the model and method used in this study, relative comparisons between values obtained from different conditions are reliable and it is more useful to compare these values to each other. The lowest acceleration values for the surface properties used (with the exception of turf, which is not a recommended playground surface) were obtained for impacts against rubber. This is due to the fact that the rubber force-displacement characteristics used in the simulations allowed for much greater displacements during the deceleration of the head.

Angular acceleration follows the same pattern as the preceding head injury parameters. Turf and rubber show sizeably lower values than the stiffer surfaces, while the tarmac and concrete values are very similar. The values obtained for angular acceleration are exceedingly high, up to an order of magnitude higher than the suggested limit of 4500rad/s^2 for rotational speeds under 30rad/s [36]. This illustrates that the absolute values are not properly validated for the type of model used in this study; only the relative values can be compared to each other, because the overall trend in the variations due to different impact surfaces is consistent, but the values predicted are too high to be physically consistent with the fact that the child actually survived the fall.

The values from stiffer surfaces (tarmac, concrete) are similar to each other, and are very similar to those obtained using a theoretically rigid surface. Consequently, using an infinitely rigid surface would be sufficient to quantify the magnitudes of the head injury parameters with a small and conservative margin of error. The difference between the results obtained from the simulations based on tarmac and concrete is negligible. Therefore, the issue of whether the impact described in this child's fall was against concrete or tarmac is not relevant.

The small changes in lower limb injury parameters observed during simulations with different impact surfaces suggest that impact absorbing surfaces do not contribute to the risk reduction of injuries to extremities or the limbs of a body as effectively as it does to the reduction of head injury.

4.2 Impact Posture

The most severe impact posture observed is the frontal impact posture, where the model strikes the ground face first. The highest injury related values calculated were those for rigid surfaces in the “front” impact posture, in which the head strikes the ground more freely. As seen from Figure 2(c), the model’s head in the frontal impact posture is the part of the dummy that strikes the ground first. This differs from the other six considered impact postures, where some other part of the body strikes the ground first, absorbing part of the impact. Another severe impact posture for head injury parameters, especially angular acceleration, is the “front45” posture (Figure 2(d)). In this position, the body accelerates towards the ground in a manner that adds a high angular velocity component to the fall. It has been seen that the actual impact posture is critical for reconstructing playground impacts, as changes in impact posture from the same drop height led to many different head injury parameter values.

There were doubts associated with this particular case as to whether the child impacted the climbing frame before striking the ground. The impact description in a case report should ideally describe if there were other strikes before the final one against the ground, since induced rotations add a level of uncertainty to the kinematics of the fall. This makes it more difficult to assess the risk of head injury accurately from a playground fall. It was observed in this study that mid-air collisions against the climbing frame have the potential to increase the magnitude of head injury parameters considerably, almost by 100%.

Computer simulations of this one particular accident allow the considerable variability of injury related parameters to be compared with respect to impact position and mid-air kinematics of the fall. Knowing the impact posture is not sufficient to properly describe a playground fall. Some knowledge of the mid-air kinematics, in the form of a description of mid-air movement and/or impacts with other objects, would ideally be required to accurately quantify the impact velocities involved in a fall.

4.3 Model Issues

While these simulation results are obtained from a 6-year-old child model, they can be extended to other age groups. Magnitudes obtained from larger or smaller models will differ in proportion to mass and body size, but the relative injury severity risk between impact surfaces or impact postures will be the same for each age group. There are differences in the relative mass of the body sections between age groups, but these do not affect the relative risks associated with impact surfaces or impact postures. While a smaller child has a relatively higher centre of gravity due to the relatively larger head mass, the impact surfaces or postures that bear more risk for a small child are the same as those that bear more risk for other age groups. Nevertheless, the risk of head injury for a younger child could be greater than for an older child.

The child head stiffness values for this study were generated using a mathematical scaling method [22], and it is not possible to guarantee biofidelity unless adequate validation of the child head stiffness model is performed. Nevertheless, the contact force values used in this study would be closer to actual child cranial stiffness characteristics than the adult skull contact characteristics. To obtain accurate contact characteristics, appropriate experiments with child

PMHS heads, and not skulls, should ideally be performed, since stiffness properties would be different because of the presence of covering tissue. Ethical issues make this challenging.

The muscles and limbs of the 6-year-old pedestrian model remain relaxed and loose during the entire event. This is seldom the case for human bodies in real situations, which tense and activate joints and muscles prior to impact. People in long duration impact situations use their limbs to protect themselves, especially their face and head (this may explain why there is a high incidence of limb injuries). Therefore, future work could usefully model muscle activation and conscious limb movement in order to establish their influence on injury risk.

5. CONCLUSIONS

The value of computer modelling, such as in the present multibody dynamics simulations, is significant during forensic investigations of accidents. The main advantage of such methods is that the range of possible outcomes that are consistent with the laws of physics and Newtonian mechanics can be examined easily. Reconstructions such as those shown on the accompanying animation graphics AVI files, can be helpful when visualising and describing a forensic scenario in a courtroom. Injuries which are manifest on a body after an accident may be identified by such means as being either primary, secondary or tertiary. The variability of alternative outcomes arising from slight differences during the sequence of an accident can also be shown. The main disadvantage of such computer simulations is the possible over interpretation of predicted results.

For the same falling posture, more compliant surfaces such as rubber do serve to reduce head injury related parameter values to a level that is considerably below that which is associated with a hard or rigid surface (30-50%). The injury related values associated with concrete and tarmac surfaces are relatively similar to each other. The orientation of a body at impact greatly influences the variability and magnitude of head injury related values.

Significantly different injury risks can be associated with falls in which the impact orientation just before impact are identical but the fall sequence either does or does not involve intermediate impact against the climbing frame prior to impacting the ground. Falls involving intermediate impacts can result in HIC scores that are more than 90% greater than unimpeded impacts against the ground.

The small changes in lower limb injury related parameters observed during simulations with different impact surfaces show that impact absorbing surfaces do not contribute significantly to reducing limb forces. This is not so for the head: energy absorbing surfaces can dramatically reduce head injury parameter values. The limited capacity of impact absorbing surfaces to reduce the risk of limb injury should not be misinterpreted to mean that playground surfaces are ineffective. Injury risk in playgrounds should be kept to a tolerable level, where the risk of serious injuries is low. Nevertheless, to completely eliminate risk from playgrounds would reduce their play value, which could, in turn, have negative consequences for the health of the child population [37].

The use of HIC has been effective in reducing head injury and its use is recommended as a tool for assessing head impact injury, but its limitations when applied to scenarios other than sagittal impacts during car crashes should be understood [38]. Additional injury criteria should be used to fully understand the possible risks associated with falls, since no one particular criterion reflects the range of mechanisms by which different traumatic brain lesions can lead to head injury [18].

In this study it has been shown that computer simulations can successfully characterise changes in injury related parameters due to different risk factors present in playground fall scenarios. They cannot completely replace physical tests, but with proper judgement, they can inform decisions and policies regarding the protective measures used in a playground. Impact simulations in a legal environment can give insight into the numerous variables involved during playground falls, and can help to analyze possible outcomes with whatever information that may be available. More work needs to be done in regards to model validation, especially on child models; feasible solutions should be found in regard to this very sensitive challenge.

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FIGURE CAPTIONS

Figure 1: Head-ground contact characteristics for adult²⁶ and scaled child (using MADYMO/SCALER²³)

Figure 2: Fall positions at impact against the ground, as analyzed in this study. Top row: (a) St45, (b) Side, (c) Front. Bottom row: (d) Front45, (e) Back, (f) Back45, and (g) Stand

Figure 3: HIC-36 values for different impact surfaces and different positions at impact. Note that these values are only for comparative purposes within this study and are not to be interpreted for establishing actual injury risk.

Figure 4: Head angular acceleration values for different impact surfaces and different positions at impact. Note that these values are only for comparative purposes within this study and are not to be interpreted for establishing actual injury risk.

Figure 5: Maximum head contact force values for different impact surfaces and different positions at impact. Note that these values are only for comparative purposes within this study and are not to be interpreted for establishing actual injury risk.

Figure 6: Head velocity against time for free fall impact and for impact against bar prior to impact against ground

Figure 7: Upper leg axial force for different impact surfaces and different positions at impact.