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Current playground surface test standards underestimate brain injury risk for children

Madelen Fahlstedt *, Svein Kleiven, Xiaogai Li

Neuronic Engineering, School of Engineering Sciences in Chemistry, Biotechnology and Health, KTH Royal Institute of Technology, Stockholm, Sweden

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ABSTRACT

Playgrounds surface test standards have been introduced to reduce the number of fatal and severe injuries. However, these test standards have several simplifications to make it practical, robust and cost-effective, such as the head is represented with a hemisphere, only the linear kinematics is evaluated and the body is excluded. Little is known about how these simplifications may influence the test results. The objective of this study was to evaluate the effect of these simplifications on global head kinematics and head injury prediction for different age groups. The finite element human body model PIPER was used and scaled to seven different age groups from 1.5 up to 18 years old, and each model was impacted at three different playground surface stiffness and three head impact locations. All simulations were performed in pairs, including and excluding the body. Linear kinematics and skull bone stress showed small influence if excluding the body while head angular kinematics and brain tissue strain were underestimated by the same simplification. The predicted performance of the three different playground surface materials, in terms of head angular kinematics and brain tissue strain, was also altered when including the body. A body and biofidelic neck need to be included, together with suitable head angular kinematics based injury thresholds, in future physical or virtual playground surface test standards to better prevent brain injuries.

1. Introduction

Playgrounds are a place with social and physical benefits for children, but also a place where many injuries occur. In Sweden, with a population of 10 million inhabitants (SCB, 2018), an estimated 18,000 children between 0 and 17 years old (YO) visit the emergency department every year due to injuries sustained at the playground (Socialstyrelsen, 2017). In US, with a population of 326 million (United States Census Bureau, 2018), the estimated annual rate of emergency department visits is 214,883 for persons 14 years or younger per year, of which 9.8% have a traumatic brain injury (TBI) (Cheng et al., 2016). Several studies have also found an increasing trend of non-fatal TBIs related to playgrounds (Adelson et al., 2018; Cheng et al., 2016). The majority of injuries are caused by falls (Adelson et al., 2018; Bierbaum et al., 2017; Mack et al., 1997; Socialstyrelsen, 2017; Vollman et al., 2009).

In an attempt to decrease the number of injuries, test standards for the impacting surface have been implemented. ASTM has introduced ASTM F1292 (ASTM International, 2017) with the purpose to decrease the risk of serious injuries and death from falls. A similar standard is available in Europe, (CSN-EN-1177, 2018) and in Australia (AS4422:2016, 2016). According to the standards, the playground surface is evaluated by dropping a 4.6 kg metal hemisphere from different heights until the critical height is found. When dropped from the critical height, the peak linear acceleration (G-max) of the hemisphere should be below 200 g and head injury criterion (HIC) below 1000. Then the playground design should be chosen so that the highest height of the playground equipment does not exceed the critical height. The critical height depends on several factors such as type of material, e.g. loose-fill materials or manufactured rubber material, the size of the grains and the moisture (Jäniskangas et al., 2017; Mack et al., 2000).

To make the test methods more practical, robust and cost-effective several simplifications are introduced. First, a hemisphere is used instead of a more biofidelic head form despite previous studies have shown that the head response is dependent on loading directions both for linear and angular kinematics (e.g. Chan et al., 2007; Gennarelli et al., 1987; Kleiven, 2005, 2003). Li and Kleiven (2018) also showed the difference between three different head impact locations (front, back and side) in playground falls with a scalable finite element (FE) human body model (HBM). Not accounting for angular kinematics is another simplification

* Corresponding author at: Hälsovägen 11, 141 57 Huddinge, Sweden.

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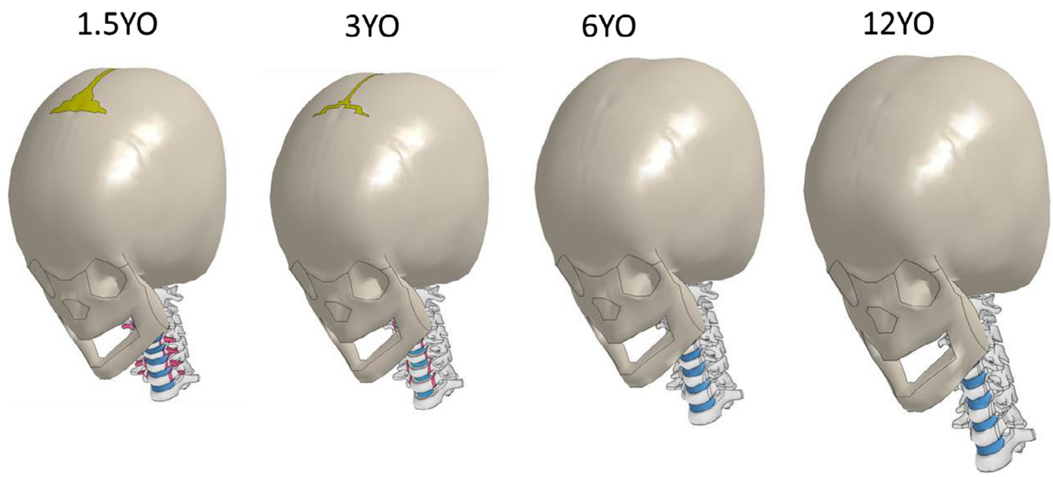

Fig. 1. The PIPER head and neck model for the 1.5, 3, 6 and 12 YO child model. The yellow areas in the skull bone for the 1.5 and 3 YO indicate the sutures while the red areas of the vertebras indicate cartilage. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

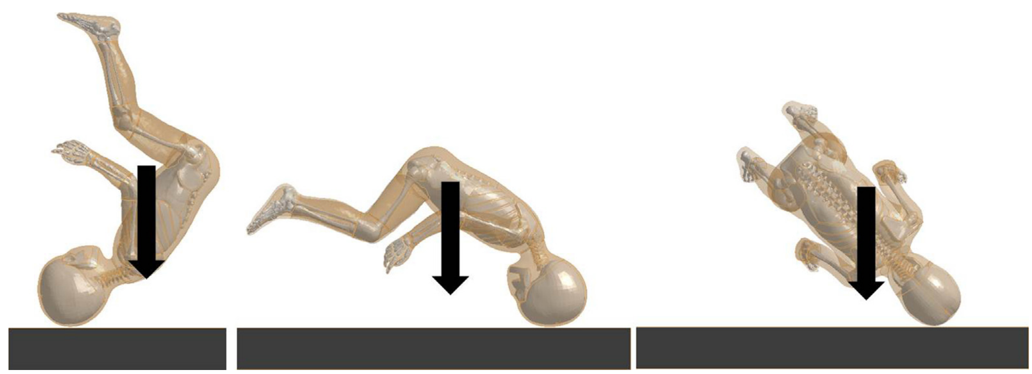

Fig. 2. The three falls (back, front, side) led to occipital, frontal and parietal impact of the head, illustrated with the 3 YO model.

with the test standards, since only G-max and HIC are measured in all three standards and additionally in the Australian standard the HIC time interval is included. Several studies (e.g. Gennarelli et al., 1972; Holbourn, 1943) have shown that brain injury correlates better with angular kinematics compared to linear kinematics, while skull fracture correlates well to linear acceleration (Kleiven, 2013). A third simplification of the test standard is that only the head is represented, excluding the body. Some previous studies have evaluated the influence of excluding the body compared to including the body in different accidents (Alvarez et al., 2014; Beusenberg et al., 2001; Fahlstedt et al., 2016; Feist and Klug, 2016; Forero Rueda, 2009; Ghajari et al., 2013; Verschueren, 2009). These studies have shown that the body influences more on the kinematics in some loading conditions and less in other situations.

All age groups of children from 0 to 18 YO are injured in playground accidents (Adelson et al., 2018; Wang et al., 2013) but a higher proportion of head and neck injuries are found among the youngest individuals (Adelson et al., 2018). The difference in anatomy between a young child compared to young adult is large, especially the head with softer skull bone with sutures as well as a higher ratio between head and body mass for the youngest children (Burdi et al., 1969).

The usage of FE models of human body has increased during the last decades in the automotive industry as an important tool to evaluate and improve safety. The advantages of these models are that they can be modelled in such details that the effect on tissue level can be studied. A previous European Union project (PIPER project; www.piper-project.org) has developed a HBM scalable from 1.5 up to 18 YO, considering both geometrical and material properties (Alvarez and Kleiven, 2018; Beillas et al., 2016; Giordano and Kleiven, 2016). Li & Kleiven (2018) performed simulations of possible playground falls and analyzed the results based on the current test standards (e.g. ASTM F1292 and EN 1177), showing that the current test methods are assuring proper protection for skull fracture but are not assuring proper protection to the brain.

The effect of the simplifications, such as using a hemisphere, only measuring linear kinematics and excluding the body, has not yet been evaluated for playground surface test standards. Therefore, the objective of this study is to evaluate these factors in the current playground surface test standard by using multilevel FE simulations to evaluate the influence on global head kinematics and head injury prediction for children of different ages.

2. Material and methods

2.1. Models

The PIPER scalable HBM (Alvarez and Kleiven, 2018; Beillas et al., 2016; Giordano et al., 2017) was used in this study. The HBM was scaled to seven ages (1.5, 3, 6, 10, 12, 14 and 18 YO) with respect to both geometrical and material properties and was positioned with the PIPER tool (v.1.0.1) prior to impact, using the same scaling and positioning as presented by Li and Kleiven (2018) (Fig. 1). Updates of the head model were also incorporated including the tentorium, trabecular skull bone, the meninges and scalp as reported earlier (Li and Kleiven, 2018).

Playgrounds of three different stiffness (hard, medium and soft) were used in this study. The surface was modelled with twelve solid elements through the thickness (total thickness was equal to 10 cm) and with the material model *MAT_SIMPLIFIED_RUBBER in LS-Dyna (LSTC, 2014). The material properties were adjusted to fit the stress-strain curve calculated from drop tests of a traditional playground rubber-composite (Huang and Chang, 2009). The G-max between simulation and experiment differed 3.3% for a drop height of 2.1 m and 4.0% for 2.5 m drop height. This was referred to as the medium stiffness playground with a critical height of 1.59 m. The playground called hard had a 5 times higher stress in the stress-strain curve and a critical height of 0.79 m. The playground called soft had 1/5 of the stress developed for the medium stiffness in the stress-train curve and a critical height of 3.26 m. More details about the playground models can be found in a previous study (Li and Kleiven, 2018).

2.2. Simulation matrix

Three falls were evaluated (back, front, side) from the same height, which led to different impact locations to the head (occipital, frontal and parietal) (Fig. 2). The impact velocity was set to 5.59 m/s perpendicular to the impacting surface, which represents a fall from 1.59 m (critical height for the medium stiff playground

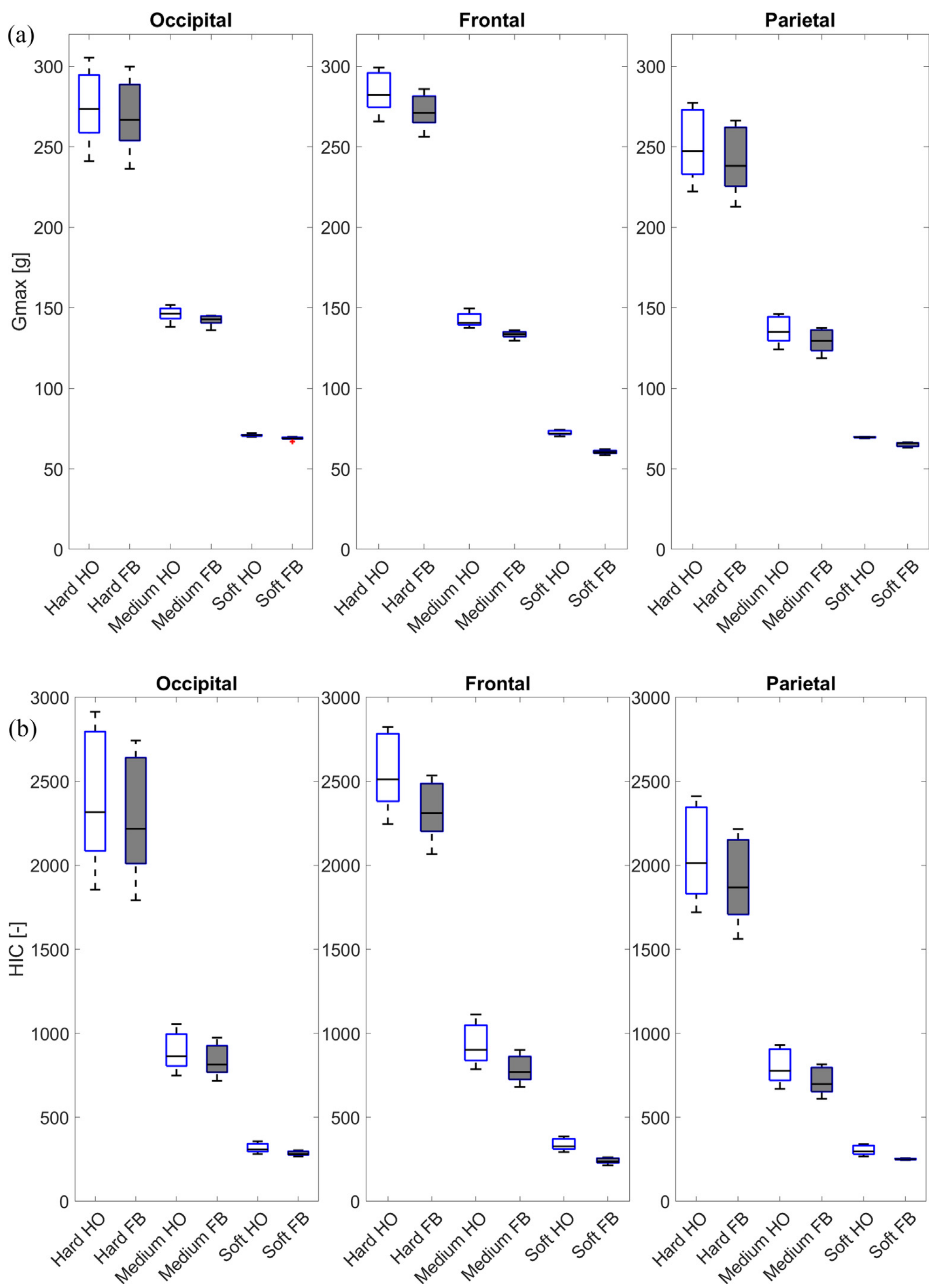

Fig. 3. The boxplots for the different impact locations (occipital, frontal and parietal) and stiffness (hard, medium and soft). The box for the full body (FB) is shown in gray and head only (HO) in white. (a) Peak resultant linear acceleration (G-max) where the outlier for the occipital soft FB simulation was found for the 3 YO model; (b) Head injury criterion (HIC).

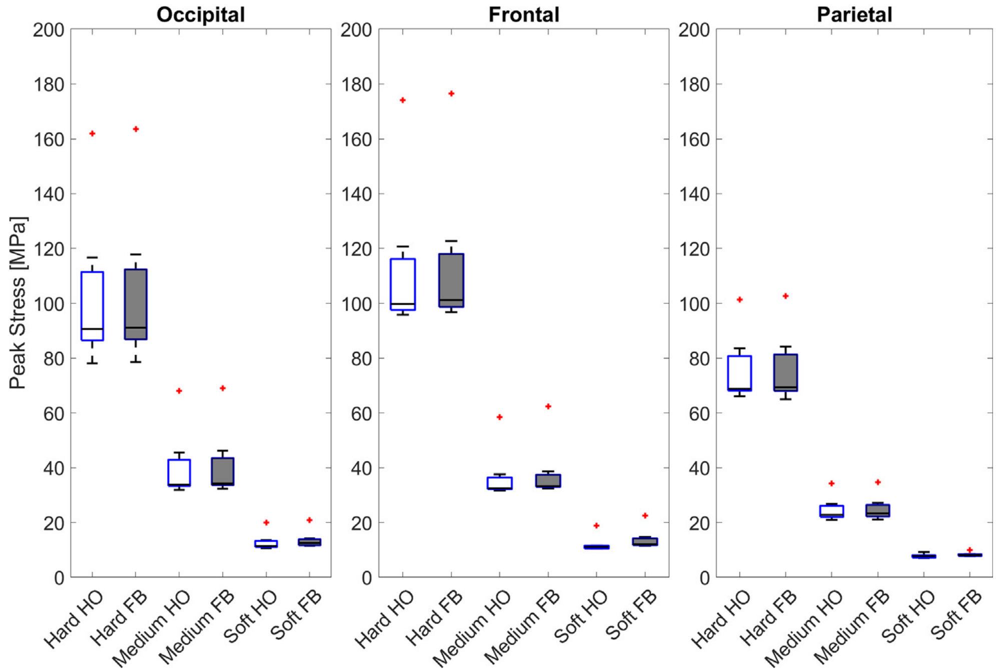

Fig. 4. The boxplot of the peak von Mises stress of the skull bone for the different impact locations (occipital, frontal and parietal) and stiffness (hard, medium and soft). The box for the full body (FB) is shown in gray and head only (HO) in white. The outliers are for the 1.5 YO model.

surface). All three impact locations were performed with the seven different age-specific models (1.5–18 YO) and the three playground surface stiffness (hard, medium and soft). All the simulations were performed in pairs, with and without the body (referred to full body (FB) and head only (HO) here after). In the HO simulations, all components below the skull base were removed except the skin and flesh of the cervical spine that had a continuous mesh with the head.

For the contact between the HBM and playground Automatic_Surface_To_Surface was used with a friction coefficient of 0.45. All simulations were performed on a multi-core Linux cluster (8.1.1 Xeon 64, 4 CPUs, LS Dyna revision 8.0.0, single precision). Mass scaling was used to achieve a time step of 0.32 μs with marginally added mass of 15 g.

2.3. Data analysis

For all simulations, the head kinematics including the resultant linear acceleration, resultant angular acceleration and resultant angular velocity were analyzed. The head kinematics was extracted from the accelerometer of the head located at the center of gravity. The linear acceleration was filtered according to the recommendations from the ASTM standard (ASTM International, 2017), which means a second order Butterworth filter with a cutoff frequency of 2077.5 Hz. The angular acceleration was filtered with a Butterworth filter with a cutoff frequency of 180 Hz. The angular velocity was not filtered. The angular velocity was included in this study since several studies have shown a correlation between brain injuries and angular velocity (e.g. Kleiven, 2005; Zhao and Ji, 2017).

The HIC value was calculated with the equation presented by Versace (1971) (Eq. (1)) with the input from the filtered resultant linear acceleration (a). Following the current playground test standards, the unlimited HIC was used in this study, which means no limitation was set for the time interval ($t_2 - t_1$).

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

The von Mises stress of the skull bone and the first principal Green-Lagrange (G-L) strain of the brain tissue were extracted from LS-PrePost (version 4.3, 64 bit, 28 March 2016) to evaluate the effect on the skull bone and the brain. These metrics have been used previously for the PIPER model to evaluate the effect on the on the skull and brain (Li and Kleiven, 2018). The 95th percentile maximum strain per element was presented following previous studies to avoid potential numerical issues (Gabler et al., 2016; Li and Kleiven, 2018; Panzer et al., 2012).

The results were presented with descriptive statistics in form of boxplots. The box in the boxplot shows the median, 25th and 75th percentile. The whiskers show the minimum and maximum value. Outliers were defined as if the value was smaller or larger than 1.5 times the interquartile range.

3. Results

The median value for G-max decreased with decreasing playground stiffness for all three impact locations for both HO and FB but only slight difference was seen between HO and FB (Fig. 3a). Similar trend was seen for HIC (Fig. 3b). The head impact locations and ages influenced the G-max and HIC. Both metrics showed a decreasing trend with increasing age for all impact locations, body configurations and playground stiffness (Figs. A1.1 and A1.2). The influence was less pronounced for the softer playground materials. The HIC time interval increased with softer playground stiffness and was dependent on impact location and age (Table A.1.1). A significant difference between HO and FB was only seen for the softest material (Table A.1.1).

A decrease of the skull bone stress with decreasing playground stiffness was also seen for all three impact locations (Fig. 4) with similar values for HO and FB. The outliers presented in Fig. 4 are the results from the 1.5 YO model. The peak stress decreased between 1.5 and 6 YO then only a small difference was seen (Fig. A1.3).

The peak resultant angular acceleration (PAA) had a decreasing trend with decreasing playground stiffness for all three impact locations and HO configuration (Fig. 5a). Meanwhile, for the FB configuration this trend was only seen for the frontal impact location. For the two other impact locations, the decrease of PAA was only seen between hard and medium playground stiffness (Fig. 5a). The influence of age showed different trends dependent on head impact location, playground stiffness and HO or FB configuration for PAA (Fig. A1.4).

For peak resultant angular velocity (PAV), the median values for FB were much higher compared to HO. An increase was seen between hard and soft playground stiffness for the FB configuration whereas there was a clear trend with decreasing PAV median value for the HO configuration (Fig. 5b). The peak values decrease with age for the FB configuration but were relatively constant or slightly increasing for the HO configuration with increasing age (Fig. A1.5).

The peak G-L strain showed a large difference between HO and FB. The peak values for the HO configuration decreased with decreasing playground stiffness but slightly increasing peak value

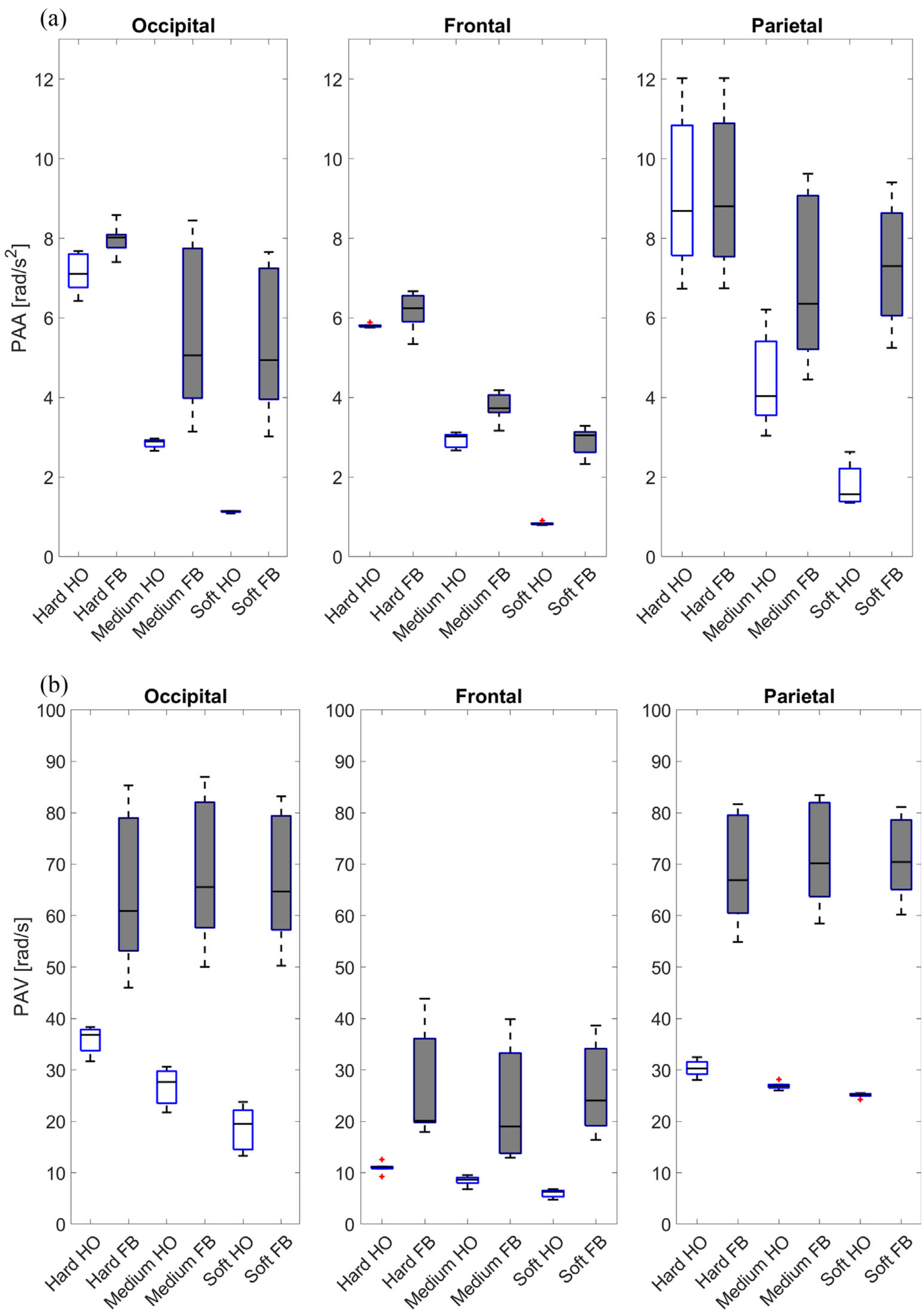

Fig. 5. The boxplots for the different impact locations (occipital, frontal and parietal) and stiffness (hard, medium and soft). The box for the full body (FB) is shown in gray and head only (HO) in white. (a) Peak resultant angular acceleration (PAA). The outlier in the frontal impact location for hard HO was found for the 10 YO model and for soft HO for 1.5 YO model. (b) Peak resultant angular velocity (PAV). The lower outlier in hard HO frontal impact was for the 3 YO and the higher outlier for the 12 YO model. For parietal impact location medium HO the outlier was found for 1.5 YO model and for soft HO for 3 YO model.

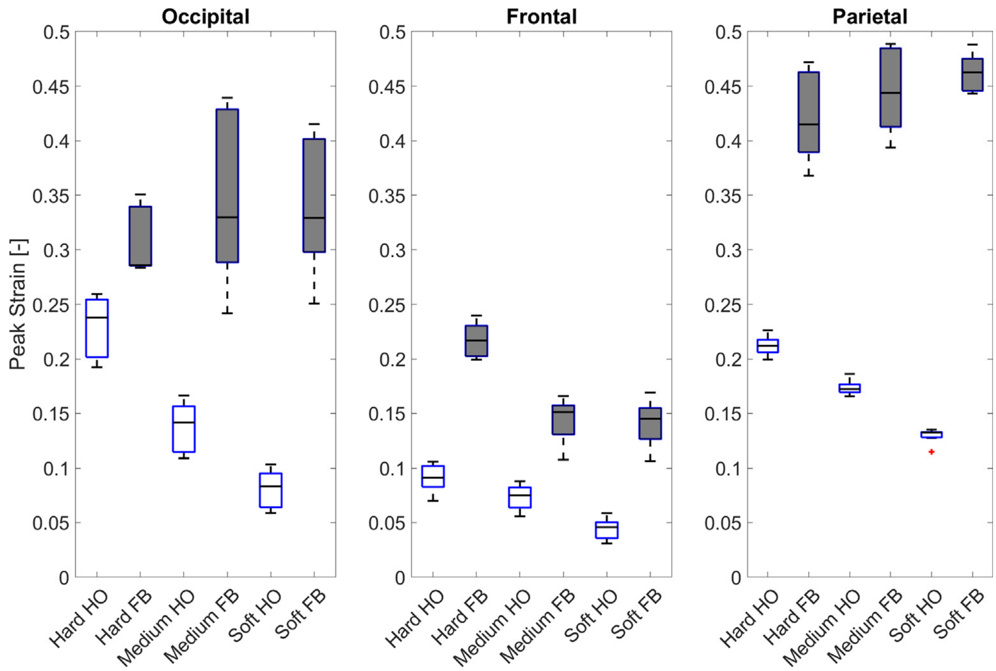

Fig. 6. The boxplot of the peak Green-Lagrange strain of the brain tissue for the different impact locations (occipital, frontal and parietal) and stiffness (hard, medium and soft). The box for the full body (FB) is shown in gray and head only (HO) in white. The outlier in parietal soft HO was found for 3 YO model. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

between hard and soft stiffness for FB configuration for occipital and parietal impact locations (Fig. 6). For the occipital impact location, the peak value increased with age for HO configuration and decreased for FB configuration (Fig. A1.6).

G-max occur at the beginning of the impact, before 15 ms (Figs. 7 and 8, illustrated with the 3 YO model) independent if the body was included or not. However, for angular acceleration and angular velocity the time for peak value was different for HO and FB (Fig. 7). The peak value for HO was before 20 ms when the head had contact with the playground. Whereas for the FB configuration the peak value occurred after 20 ms when the body had more time to influence the kinematics and start to interact with the playground.

4. Discussion

The objective of this study was to investigate how the simplifications introduced to the current test standard to make it practical, robust and cost-effective influences the performance of a playground surface (ASTM F1292, AS 4422:2016 and EN 1177). This includes the usage of a hemisphere, only measuring the linear kinematics and excluding the body and neck. Head angular kinematics and brain tissue strain were both influenced by head impact location and the body (Figs. 5 and 6). The angular kinematics and brain tissue strain were underestimated by the simplification of excluding the body as in the current test standards. Simplifications in standards are mainly performed to make the standard practical, robust and cost-effective. However, the simplifications need to be balanced against optimizing the injury prevention and above finding urges a more stringent requirement in the testing standard to protect the brain. Linear kinematics, on the other hand, was only slightly influenced by the head impact location for both HO and FB (Fig. 3). A slightly higher values for HO compared to FB were found, which suggests that the current test method is conservative when using linear kinematics if we assume FB better represents the real situation.

The aim of the test standards of playground surfaces is to promote safer surfaces and excluded inferior design. In this study three different designs with different stiffness were evaluated. As mentioned above the linear kinematics and skull bone stress were only slightly influenced by the simplifications of the test standards but the ranking of performance of the three playground materials was not changed. Independent of head impact locations or including/excluding the body the results suggested larger prevention with softer impacting surface. However, the same ranking between the three designs were not found for PAA, PAV and peak first principal G-L strain, where the ranking of performance were more scattered, indicating the simplification of excluding the body and neck played a significant role. In all three impact locations, the HO configurations predicted better prevention for more compliant impacting surfaces although the impact locations influenced the peak value predictions. While for the FB configuration, the decrease of median peak value with decreasing surface stiffness from hard to soft stiffness was only seen for PAA in the frontal impact location. In other cases, e.g. PAA for the occipital impact location, the decrease in peak value was only seen between hard and medium stiffness. There were also cases, e.g. peak strain for the parietal impact location, where the median peak value increased with decreasing surface stiffness.

The age effect showed large influence on the stress of the skull bone for the youngest individuals. This can be explained by the much more compliant material properties of the skull bone in the 1.5 YO (Fig. A1.3). For G-max and HIC the decrease of peak value with age was more linear (Figs. A1.1 and A1.2). The simulations performed in this study are limited to radial impacts, and the angular kinematics of the head in the HO simulations was induced by its contact with the playground. After the head lost contact with the impacting surface only small changes of the angular velocity was seen. While in the FB simulations, the angular kinematics was influenced both by the contact between the head and playground and the motion of the body. Therefore, the angular velocity continued to increase in the FB configuration after the head had lost contact with the impacting surface.

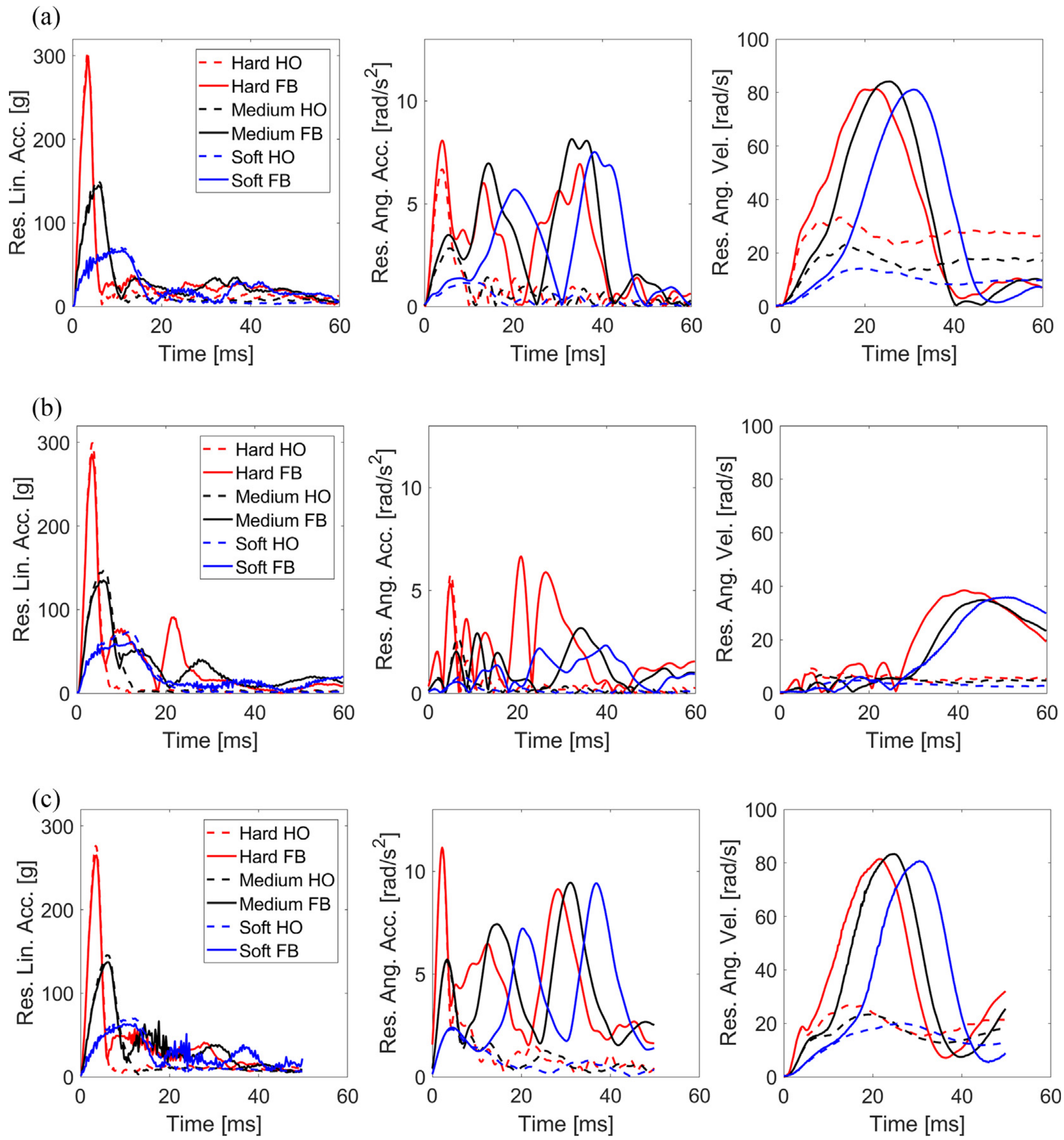

Fig. 7. The kinematics of the 3 YO model with the different stiffness for the three impact locations, (a) occipital; (b) frontal; (c) parietal. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

The current test standards are promoting softer impacting surface until the material is bottoming out, since only the linear kinematics is measured, which gives an optimization of the playground surfaces against skull fracture. This is also mirrored in the epidemiological studies. Mitchell et al. (2007) found a decrease of severe head injuries including skull fractures and intracranial injuries between 1992 and 2004 in Australia while several recent studies (Adelson et al., 2018; Cheng et al., 2016) have shown an increasing trend of concussions and other closed head injuries over the last decade. The results in this study show that an improvement of the current test standard by including a more biofidelic head form and measure the angular kinematics is probably not enough to optimize the playground surface test standard against brain injuries. Since excluding or including the body are giving different ranking of the three impact surfaces evaluated and underestimated the risk of brain injuries when excluding the body. The question is what alternatives are possible to improve and optimize the current test standards. Previous studies involving helmet test standards have proposed to modify the inertia and center of gravity of the head form to take into account the influence of the body (Feist and Klug, 2016; Ghajari et al., 2013). In the consumer test of vehicle safety (new car assessment programme NCAP) whole body dummy models for the occupant safety are used (EuroNCAP, 2017a) and for pedestrian safety FE HBMs has been introduced (EuroNCAP, 2017b). The HBMs in the EuroNCAP tests need to fulfill certain certification before they can be used, which is based on results from post mortem human subject (PMHS) tests. For pedestrian impacts there are several PMHS tests available (e.g. Forman et al., 2015; Subit et al., 2008), which the models can be evaluated against. Experimental data of falls, especially for children, are rare. One alternative would be to use whole body anthropometric test devices such as dummies in the standards for playground surfaces testing but they also require experimental data to be evaluated to know more of the suitability for the usage as substitute for the child.

The lack of experimental data of children and fall accidents are also one of the limitations of this study. The PIPER HBM has been evaluated on component level for the brain, cervical spine, trunk and lower extremities (Alvarez and Kleiven, 2018; Beillas et al., 2016; Li and Kleiven, 2018) with results mainly laying within the corridors. The whole body evaluation has been focusing on side and frontal collision with validation against sled experiments (Beillas et al., 2016). Another limitation of the HBM is that only

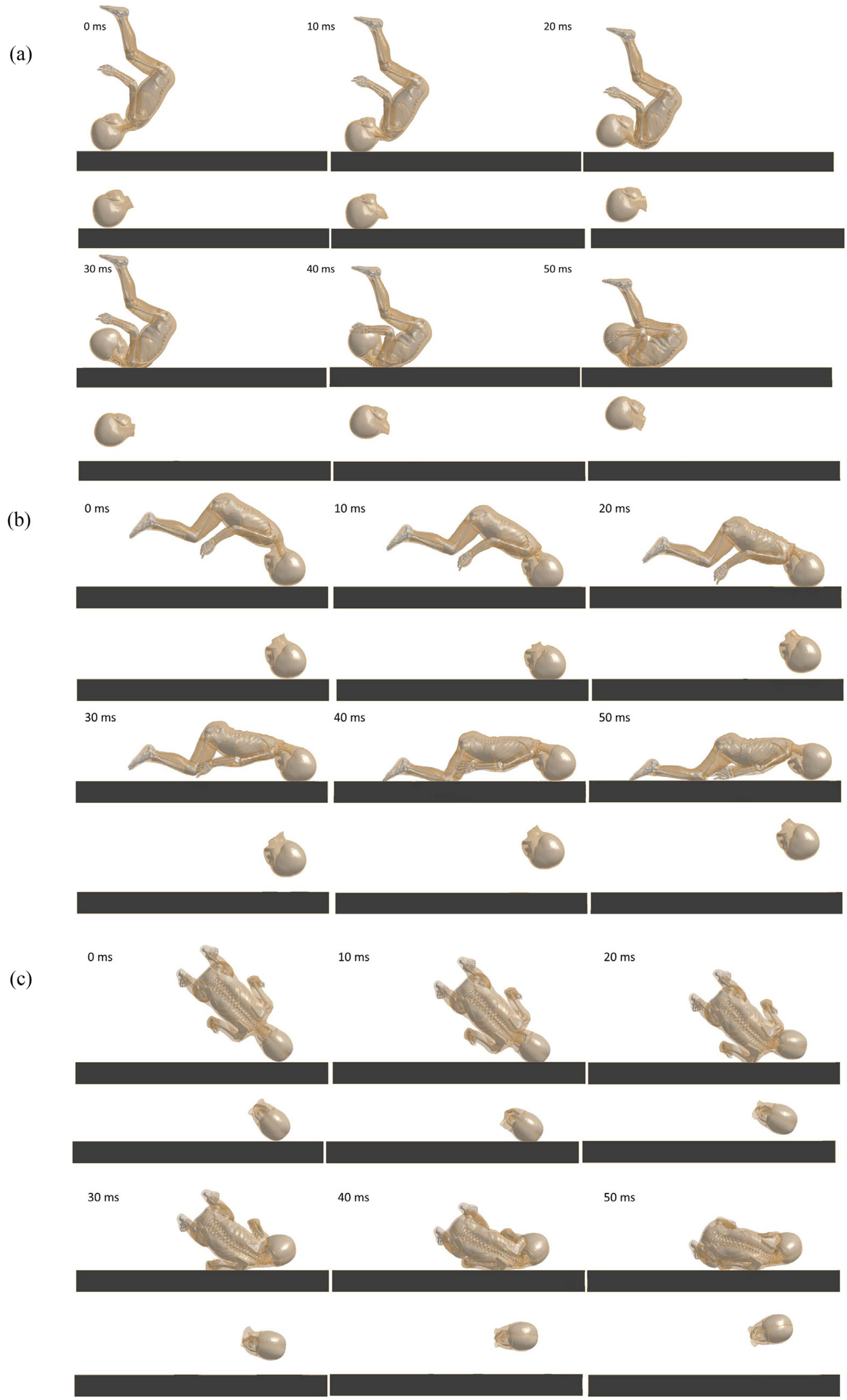

Fig. 8. Illustration of the fall for the 3 YO model with the medium stiff for FB and HO for every 10 ms, (a) occipital; (b) frontal; (c) parietal.

passive muscles have been used. Active muscles can influence the results, which has been shown in other studies (Alvarez et al., 2014; Brolin et al., 2005; Fahlstedt et al., 2016; Jin et al., 2017). But the question is how much time it takes for a human to activate muscles in a fall and how much forces are generated that may affect the head kinematics.

Adelson et al. (2018) showed that the incidence of playground related injuries was higher for children between 5 and 12 YO, but concussion and closed head injuries had highest incidence among the youngest children, equal or less than 4 YO. In our study, seven different age-specific models were evaluated where the geometry and material properties were changed with age compared to the test standard where one hemisphere is used. The G-max, HIC and peak stress of the skull bone had highest peak values for the youngest children independent of impact location or including/excluding the body but the age effect was influenced by the stiffness of the impacting surface. For PAA, PAV and peak strain the youngest model did not always produce the highest peak values. The results were influenced by the impact location, injury metrics and body configuration. The age effect needs to be further investigated to understand if different test apparatus are needed.

This study is focusing on head injuries but other injury types are also common in playground falls, such as fracture in the upper extremities (Adelson et al., 2018). Other injury mechanisms are found among these injuries, which should also be taken into account when discussing improvement of the standards for impacting surface in playgrounds. Eager and Hayati (2019) have discussed including measurements of impulse force and bounce to better evaluate the impacts also for long-bone fractures. To improve and optimize the test standards for playground impacting surface more research is also needed to understand the impact situation further. Questions around how a typical fall looks like and the worst case scenario could be explored more as a complement to previous study (Foust et al., 1977).

This study shows that the simplifications of the current test standards of playground surfaces influence the performance for both linear and angular kinematics as well as skull fracture risk and brain injury risk. However, the simplifications have a larger effect on the angular kinematics and brain injury predictions. The performance of the playground is influenced by the impact location of the head, whether the body is included or not and the age of the human model. The ranking of material with different stiffness was influenced by the body configuration (HO/FB) for the angular kinematics and peak first principal strain. The results suggested that the current test standard is not optimized for preventing brain injuries and the simplifications need to be carefully investigated to optimize the standards towards prevention of different types of injuries.

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Conflict of Interest

We have no conflict of interest.

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2019.03.038.

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