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Mechanics of Soccer Heading and Protective Headgear

March 28, 2018

This book is dedicated to my dearest family,
Nur Aqilah, Hana Ayesha, Haifa Aaleya, and
Huda Awfaa.

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Contents

Chapter 1
Concussion in Soccer

Abstract This chapter covers the review of literature that motivate this work. This includes a brief overview of sports-related head injuries particularly in soccer, and the efficacy of protective headgears in mitigating the impact due to soccer heading.

1.1 Sports-related Concussion

Although exercise and sports are regarded as healthy activities, they can lead to physical injuries. The worst physical injury that can be sustained by an athlete is the head or brain injury since the brain controls both the mental and physical performance. Concussions are a surprisingly common occurrence in sports and recreational activities. Concussions may result from a fall or from players colliding with each other, the ground, or with obstacles such as a goalpost. High school athletes suffer thousands of concussions every year, most often in football, ice hockey, and soccer.

It was reported that more than 200,000 cases of sports-related concussions occur in the United States annually [1]. The American Association of Neurological Surgeons listed 20 sports and recreational activities with the highest number of head injury cases treated in the U.S. hospitals in 2009 as shown in Figure 1.1. Amongst the sports with high cases of head injury are cycling, American football, baseball and softball, basketball, water sports, soccer, etc. A statistical data from Hootman et al. [3] suggests that over the past decades, the annual sports-related concussion rate shows an increasing trend. Further, female athletes were found to sustain more concussions than male athletes [4, 5].

Furthermore, sports-related concussion in children has always been a concern. A study showed that more than 53.4% and 42.9% of serious head injuries in children of age between $10-14$ and $15-19$, respectively, were sports-related [6]. It has also been found that the outcomes of concussion are associated with age and sex, where high school athletes were found to perform worse in several neuropsychological tests such as verbal and visual memory compared to college athletes [7]. In addition,
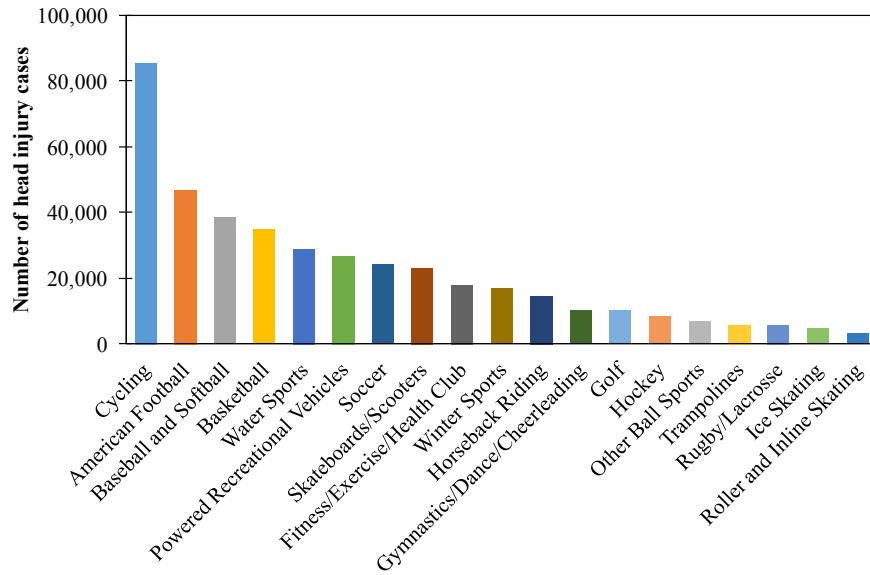
Fig. 1.1: Sports-related head injury cases treated in the U.S. hospitals in 2009 [2].

a recent study by Stamm et al. [8] showed that an exposure to repetitive head impacts during the critical neurodevelopment phase (prior to the age of 12) may result in permanent changes of the brain microstructure. This might affect later-life mood and behaviour that are associated with cognitive impairment.

In many cases, the concussions were not reported, or even worse, Delaney et al. [9] found that some of the athletes did not realise that they had suffered a concussion. Studies have demonstrated that the risk of sustaining a concussion is higher for those who have had a previous concussion [10]. This suggests that the cumulative effect of concussions might lead to a more serious brain injury. The worst side of cumulative concussion is called the 'second-impact syndrome', a condition in which the brain swells rapidly and catastrophically after a person suffers a second concussion before symptoms from an earlier one have subsided. The most significant difference of sports-related concussion is that the athlete will most likely be returning to play and will potentially be exposed to another head impact.

Serious head injuries might occur during sporting activities, which could be fatal. In 2002, a 59-year-old England's soccer legend, Jeffrey Astle collapsed at his daughter's home, and died shortly after that. The coroner has concluded that the cause of the death was a degenerative brain disease. Astle, who was a frequent soccer ball header might have sustained repetitive minor traumas as a result of soccer heading manoeuvre. This might be the reason behind his death. Patrick Grange, a soccer player from a leading soccer club in America, Chicago Fire, died at the age of 29. He was very popular for his heading ability. Researchers from Boston University found that his death was due to the chronic traumatic encephalopathy (CTE),

a degenerative brain disease associated with repetitive head injury. CTE frequently occurs in players experiencing sports-related head injury. In 2014, a young Australian cricketer, Phillip Hughes was struck by a cricket ball on the neck during a match. This has led to a vertebral artery dissection that led to a subarachnoid haemorrhage. He was placed under induced coma; two days later, he died. These few examples show that head injuries during sporting activities need to be taken into serious consideration; a few unfortunate cases might result in casualties.

1.2 Concussion in Soccer

Soccer is the most popular sports in the world with an active involvement of more than 4% of the world's population [11]. This game is not traditionally associated with head injuries. Nonetheless, it was reported that soccer players can sustain as many concussions as the American football or ice hockey players [12, 13]. Concussions in soccer could occur due to collision of the head with other player's head, elbow, knee, ground, ball and goal post. A previous study stated that the concussion accounts for approximately 22% of the overall soccer injuries, which rate shows an increasing trend [14].

Delaney et al. [15] reported that more than 60% of the university soccer players had sustained concussion during a year of participation. However, only 19.8% of them realised that they had suffered a concussion. The study also found that more than 80% soccer players had sustained multiple concussions. This study shows that soccer players experience a significant number of concussions; soccer can no longer be recognised as safe from head injuries.

1.3 Soccer Heading and Concussion

Soccer is a unique sport, in which the players are permitted to use all body parts except for the hands to control and pass the ball to the teammates. The head is also used during passing and even to score goals, and this manoeuvre is termed 'heading'. This purposeful use of the head in soccer has raised concerns as to whether it could lead to brain trauma injury. A soccer player could be subjected to six to seven occasions of heading during a game [17] with an estimation of approximately 800 headings a year [18]. This, however, does not include the number of headings sustained during training sessions. Thus, a professional soccer player might experience up to more than a thousand headings annually.

Many studies have linked purposeful heading in soccer to brain trauma injury, similar to that found in mTBI. Both amateur and professional soccer players were evaluated in the past decades. This was done through a series of neuropsychological tests [19, 20, 21, 22, 23]. These tests assess the neurocognitive performance of the players in several aspects such as planning, memory, attention, visual and verbal.

It was found that soccer players had scored poorly in the tests as compared with the non-soccer players [20, 23]. Frequent headers obtained even lower scores when compared with the non-headers [19, 21, 22]. Thus, the number of headings was reported to be inversely proportional to the neurocognitive performance of a soccer player [20].

In addition to the tests, a tablet-based application was developed and used to evaluate the cognitive effect of soccer ball heading [24]. Results obtained demonstrated that soccer players were significantly slower than control group in voluntary response task. This suggests that soccer ball heading could lead to alterations of cognitive function that are similar to mTBI. This study introduces an easier method in assessing the cognitive functions that can be applied not only to soccer players, but also to athletes involve in other contact sports.

Researchers have also used a more advanced method known as diffusion tensor imaging (DTI) in assessing the neurocognitive performance of soccer players. DTI is an advanced magnetic resonance imaging technique that offers detection of microscopic changes in the brain's white matter. It measures the movement of water molecules along the nerve fibres in the brain (axons) that is known as fractional anisotropy (FA) [25]. Researchers have attempted to assess both amateur and professional soccer players using this technique [26, 25]. These studies have found abnormalities in the white matter of the frequent headers similar to that found in patients with mTBI. In addition, the former study found compelling evidence that exceeding a threshold level of 885 to 1,550 headings a year results in significantly lower FA values that is associated with cognitive impairment in patients with TBI. Lipton et al. [25] also stated that some people might be at higher risk of sustaining brain injury despite having an exposure to soccer heading below the threshold level.

Besides the neurocognitive test and brain imaging, the serum concentration of the biomechanical markers for brain damage was also used to evaluate the adverse impact of soccer heading. Amongst the biomechanical markers assessed were the S100-B, neuron-specific enolase (NSE), nerve growth factor (NGF) and brain-derived neurotrophic factor (BDNF). Stålnacke et al. [27, 28] detected an increase in both S100-B and NSE biomarkers following competitive soccer game; the increase of S100-B was associated with the heading frequency. A significant increase in S100-B serum was also reported by Mussack et al. [29] after a soccer heading activity. In addition, Bamac et al. [30] recorded an increase in serum concentrations of NGF and BDNF as a result of soccer heading manoeuvre.

On the other hand, there were also studies that found no association between soccer heading and neurocognitive impairment, for instance a study by Guskiewicz et al. [31] on collegiate soccer players. A comparison between the results of neuropsychological tests between soccer players, non-soccer players and college students showed no statistical significance. Nonetheless, this study did not measure the exposure to soccer heading. Participation in soccer, without being exposed to soccer heading or head collision is undoubtedly safe. Thus, the results of Guskiewicz et al. [31] is true as far as only being involved in soccer game is concerned. It would be interesting to compare the test results of frequent soccer ball headers with the less frequent counterparts or the non-athlete. Moreover, a recent study by Chrisman

et al. [32] also found no relationship between head impact exposure with cognitive deficit. The data was recorded during a weekend youth soccer tournament. The fact that each subject was assessed less than a week before and after the tournament shows that head impacts in soccer do not have an immediate adverse effects on the brain. The real concern, however, is the cumulative effect that might lead to cognitive impairment in 20 years time.

The studies discussed in this section demonstrated that heading in soccer might be harmful to the brain. Although the cognitive impairment might also be caused by other head impact cases such as head-to-head or head-to-elbow impacts, the detrimental effects of head-to-ball impact cannot be simply ruled out. A single heading might not be dangerous, but the cumulative effects of the repeated blows are the main concerns. A proper heading technique could reduce the injury risk; but there are a lot of external factors involved when performing a soccer heading manoeuvre such as the opponent players or the ball spin that are hard to be controlled. These factors make it difficult to execute a proper heading every time. Banning heading from the game is the ultimate solution; nonetheless it does not seem ideal. A protective headgear that can attenuate the impact from the ball during heading could be a more practical solution.

1.4 Soccer Headgear

The increase of head injury cases in soccer has spurred the introduction of several headgears designed specifically for the head protection during a soccer game. Amongst the notable soccer headgears are the Full90, ForceField, DonJoy Hat Trick and Storelli Exoshield Headguard (Table 1.1). These headgears are made of impact-absorbing foams such as the polyurethane and polyethylene. The designs vary from one manufacturer to another. However, the most important question is how effective are these headgears in protecting the brain during a soccer heading manoeuvre?

(a) Full90   (b) ForceField   (c) Storelli Exoshield

Fig. 1.2: Commercial soccer headgears.

Studies have been conducted to evaluate the effectiveness of these headgears. Amongst the earliest study was by Naunheim et al. [33]. Four headgears were stud-

Table 1.1: Commercial soccer headgears and their respective manufacturers.

Model	Manufacturers	Country
Full90 Premier	Full90 Sports Inc.	San Diego, CA, USA
ForceField FF Protective Headgear	ForceField FF (NA), Ltd.	USA
DonJoy Hat Trick	DJO Global, Inc.	Vista, CA, USA
ExoShield Headguard	Storelli Sports Inc.	New York, USA

ied: Soccer Docs, Kangaroo, Head Blast and Head'r. These headgears were fitted to a standard anthropomorphic headform instrumented with a triaxial accelerometer at the centre of gravity of the headform. A soccer ball was launched at three different velocities: 9, 12 and 15 m/s; the linear head acceleration due to the ball impact was measured. This study concluded that all headgears tested were not effective in reducing the impact during the simulated soccer heading.

In addition to the headform, soccer headgears were also tested using a force platform [34]. A soccer ball was launched at the force platform at a speed of 15.6 m/s. Three soccer headgears were tested: Headers, Head Blast and Protector. The peak impact force were recorded, the time to reach the peak force and the impulse were calculated. The headgears were found to reduce the peak impact force, but only up to 12%. Nonetheless, the authors recommended further studies to investigate the effectiveness of these headgears in reducing the risk of sustaining TBI during soccer heading manoeuvre.

Withnall et al. [35] tested three headgears: Full90, Head Blast and Kangaroo Soccer Headgear on human subject for low-speed ball impact and on the Hybrid III crash test dummy for the high-speed ball impact. In the low-speed test, the human subject was instrumented with a mouthpiece instrumented with two accelerometers; the ball was projected to the subject at two speeds: 6.4 and 8.2 m/s. In the high-speed test, the ball was launched at 10, 20 and 30 m/s. Besides ball-to-head impact test, the authors also performed a head-to-head impact test using two dummy headforms. The results demonstrated that for the ball-to-head impact, all headgears tested were unable to reduce the head accelerations and HIP_{max}. Nonetheless, in the head-to-head impact test, the headgears provided an overall of 33% impact protection. This suggests that the headgears are only effective in a collision with a hard, non-deformable object such as the head or a goal post, but provide no protection in an impact with a soft, deformable object such as a soccer ball.

Besides experimental work, a computer simulation was also used to investigate the effectiveness of a soccer headgear. Lehner et al. [36] used a detailed human head-neck model to simulate soccer heading manoeuvre and evaluate the effectiveness of the Full90 headgear in reducing the risk of brain injury. The ball speed simulated was between 4 to 30 m/s. The linear and angular accelerations were measured, and the HIP_{max} was calculated. The results showed that the headgear provided less than 5% reduction of the HIP_{max}, which suggests that wearing the headgear during soccer heading manoeuvre provides no measurable attenuation.

Very recently, Elbin et al. [37] conducted an experiment to evaluate the neurocognitive performance following a bout of soccer heading in athletes wearing soccer headgear. Twenty-five subjects were divided into two groups: the headgear and no headgear group; the subjects in the headgear group wore the Full90 headgear during the experiment. The assessment of the neurocognitive function was achieved using the Immediate Post-Concussion Assessment Cognitive Testing (ImPACT), a widely used test for concussion management. The ball was launched at a speed of 22.35 m/s; each subject headed the ball 15 times within a 15-minute period. Each subject was administered a neurocognitive test before and immediately after the experiment. The results revealed the the headgear was not able to attenuate the subtle neurocognitive effects of acute soccer heading. On the contrary, the authors found that the no headgear group performed better than the headgear group in terms of verbal memory and reaction time.

1.5 Summary

This chapter discusses the literature that motivates the present work. Statistics show that the number of TBI cases shows an increasing rate, and it is not uncommon that sporting activities is one of the main causes. The most concerning fact, however, is that most of the sports-related head impacts are subconcussive that most of them went unidentified. Studies have associated the frequency of soccer heading with brain injury. Evaluation of soccer players' brain presented a compelling evidence to link the cumulative blows to the head due to soccer heading to mTBI. Soccer headgears were introduced to reduce the risk of sustaining concussion. However, studies demonstrated that the headgears did not provide any measurable protection against soccer heading impact. With the increasing evidence of the adverse effect of soccer heading to the brain and the inability of the commercial headgears to provide a significant protection, there is a need to further understand the mechanics of soccer heading manoeuvre to reduce the energy transmitted to the brain. Nevertheless, reducing this would also reduce the energy transferred to the ball, which could be a displeasure amongst adult players. Even so, this study is urgently needed to protect the minors and children from a potential risk. Therefore, the impact-absorbing foam that can be used to mitigate the impact due to soccer heading needs to be investigated.

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Chapter 2
Soccer Ball Finite Element Model

Abstract This chapter describes the development of the soccer ball FE model and its validation. The geometry of the ball and the material properties of each layer of the ball model are presented. In addition, the pressurisation of the ball is also described. The model was validated against a more advanced soccer ball model.

2.1 Introduction

In soccer, the ball itself is one of the key equipment. The initial vulcanised soccer ball was introduced more than 150 years ago by Charles Goodyear, before he invented the first inflatable soccer ball in the year 1862 [1]. In a soccer game, the ball undergoes a lot of contacts with the players, the ground as well as the goal post. Researchers have used numerical methods to study the mechanics of soccer ball impact. Nagurka and Huang [2, 3] used mass-spring-damper model to study the ball impact while Price et al. [4, 5, 6, 7] developed advanced finite element model of soccer ball for the same purpose in addition to finding ways to improve soccer ball design.

This work aims to analyse soccer heading by means of FE analysis. Hence, using an advanced soccer ball model such as that of Price et al. [4, 5, 6, 7] might result in a very high computational cost when it comes to soccer heading simulation. Thus, this chapter discusses the development of a simplified soccer ball model, but with the target to retain the accuracy of the prediction compared to the more advanced model.

2.2 Geometry and Meshing

The ball was modelled using a composite sphere shell. It has a diameter of 220 mm, which corresponds to a standard size 5 soccer ball. It comprises of two layers; an

inner latex bladder and an outer composite panel. The former layer is 0.8 mm thick while the latter is 2.2 mm thick.

To obtain uniform mesh, the ball was partitioned into four sections as shown in Fig. 2.1. This was achieved by creating two datum planes; $x - y$ and $y - z$. The ball

Fig. 2.1 Partitioning of the sphere shell into four sections, creating a spherical octahedron.
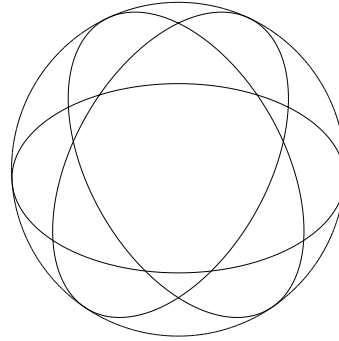
was meshed using linear quadrilateral elements of type S4R in Abaqus/CAE with structured mesh technique. Fig. 2.2 depicts the structured uniform mesh of the ball model. Altogether, the ball model contains 2,904 elements.

Fig. 2.2 Structured quadrilateral mesh of the ball model.
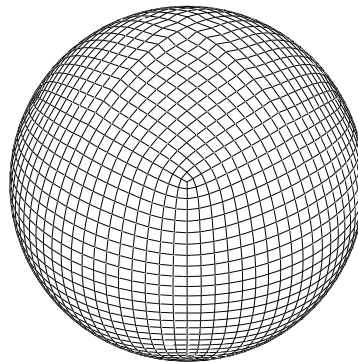
2.3 Material Properties

As mentioned in the previous section, the ball model consists of two layers. The material properties of each layer was extracted from the tensile test result of Price et al. [4] as shown in Fig. 2.3. The graphs were digitised into stress-strain data,

Fig. 2.3 Tensile test data of the inner latex bladder and outer composite layer of Price et al. [4].
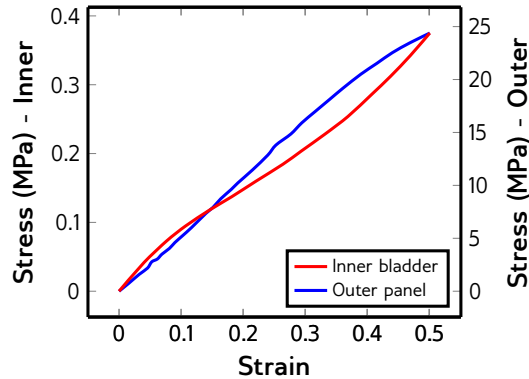
which were then input into the software. Hyperelastic material model with reduced polynomial strain energy potential was used for both layers. The evaluation of the tensile response data of both layers shows that the inner bladder is best described by the 4th order strain energy potential, while the outer panel by the 5th order strain energy potential.

The density of the inner bladder and the outer panel was defined as 1,175 kg/m^3 and 900 kg/m^3, respectively. This results in a total mass of 0.444 kg, which conforms to the mass of a standard size 5 soccer ball. In addition, a stiffness-proportional damping coefficient was defined for the outer panel to provide energy losses throughout the impact. This was done to increase the accuracy of the impact response of the soccer ball model. The damping coefficient cannot be physically determined. From a parametric study, it was found that the damping coefficient of 0.0003 provides the best agreement with the published experimental results.

2.4 Ball Pressurisation

Pressurisation of the ball was implemented using the *surface-based fluid cavity* technique instead of the *element-based hydrostatic fluid cavity* technique executed by Price et al. [4]. These modelling techniques are used in Abaqus/CAE to define the fluid-filled structures. The former technique supersedes the element-based hydrostatic fluid cavity capability in terms of functionality and eliminates the need for

fluid or fluid link element definition [8]. It is assumed that the cavity of the ball is filled with fluid that possesses uniform properties and state.

A cavity reference node, which has a single degree of freedom, was defined at the centre of gravity of the ball to represent the pressure inside the cavity. The node was coupled to the inner surface of the ball as shown in Fig. 2.4. The fluid cavity prop-

Fig. 2.4 Pressurisation of the ball using surface-based fluid cavity. A node (cavity point) was created at the centre of gravity of the ball and coupled to the inner surface of the ball (cavity surface).
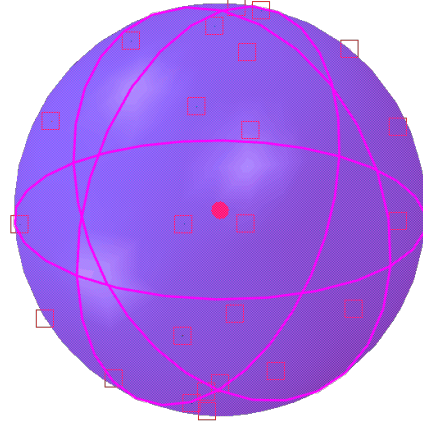
erties require the definition of heat capacity at constant pressure. This was defined using a polynomial form based on the Shomate equation according to the National Institute of Standards and Technology, as follows:

$$\tilde{c}_p = \tilde{a} + \tilde{b}(\theta - \theta^Z) + \tilde{c}(\theta - \theta^Z)^2 + \tilde{d}(\theta - \theta^Z)^3 + \frac{\tilde{e}}{(\theta - \theta^Z)^2}, \qquad (2.1)$$

where $\tilde{a} = 28.110$, $\tilde{b} = 1.967 \times 10^{-3}$, $\tilde{c} = 4.802 \times 10^{-6}$, $\tilde{d} = -1.966 \times 10^{-9}$, and $\tilde{e} = 0$ are gas constants. The ideal gas molecular weight is taken as 0.0289.

One step was created and exclusively dedicated to the ball pressurisation. Pressurisation was performed gradually using *smooth step* amplitude function since an abrupt increase in pressure inside the cavity will result in the explosion of the ball. At the end of the step, the ball is pressurised to the desired level. To avoid further increase in pressure, the pressurisation was deactivated in subsequent steps.

2.5 Ball Impact Simulation

To perform the ball impact simulation, a rigid wall was modelled using planar shell feature. The ball and the wall was assembled. The ball was placed very close to the

wall to reduce the computation time as shown in Fig. 2.5. A reference node was

Fig. 2.5 The ball was assem-
bled with a rigid wall. It was
placed very near to the wall to
reduce computation time.
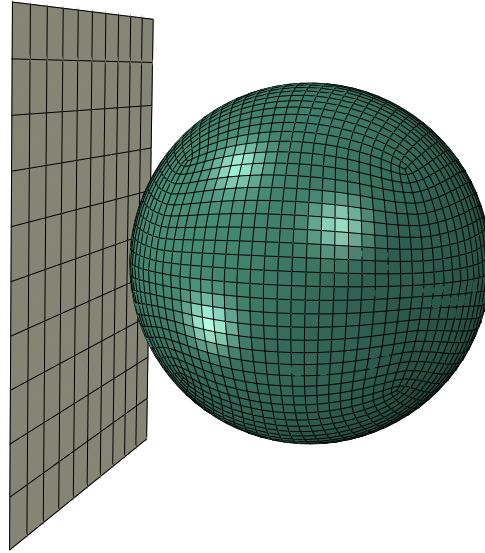
created at the centre of gravity of the ball. It was coupled to the outer and inner
surfaces of the sphere shell using the *structural coupling* function in Abaqus/CAE.
Then, a velocity boundary condition (BC) was defined at the reference node, which
represents the inbound ball velocity. The centre of gravity of the rigid wall was fixed
in all translational and rotational degrees of freedom.

To ensure the ball impacts the rigid wall, the velocity BC has to be deactivated
before impact. Hence, a new step was created for impact. Altogether, there were
three steps created for this simulation; *pressurise*, *ball velocity*, and *impact*. In addi-
tion, a *General Contact* was defined to ensure a proper contact between the ball and
the rigid wall. The coefficient of friction was arbitrarily chosen as 0.5 since it was
found to have negligible effect on the impact response [9].

2.6 Model Validation

The aim of this work was to develop a simplified soccer ball model that is validated
against published dynamic impact test data and a more advanced model of Price
et al. [4]. Therefore, the dynamic ball impact experiment of Price et al. [4] was
simulated. The ball was inflated to 0.9 bar and the inbound velocity was defined as
11, 15, 20, 22, and 28 m/s. The parameters measured from the simulation were the

coefficient of restitution (COR), contact time, and longitudinal deformation. Fig. 2.6, 2.7 and 2.8 shows the comparison between the two models.
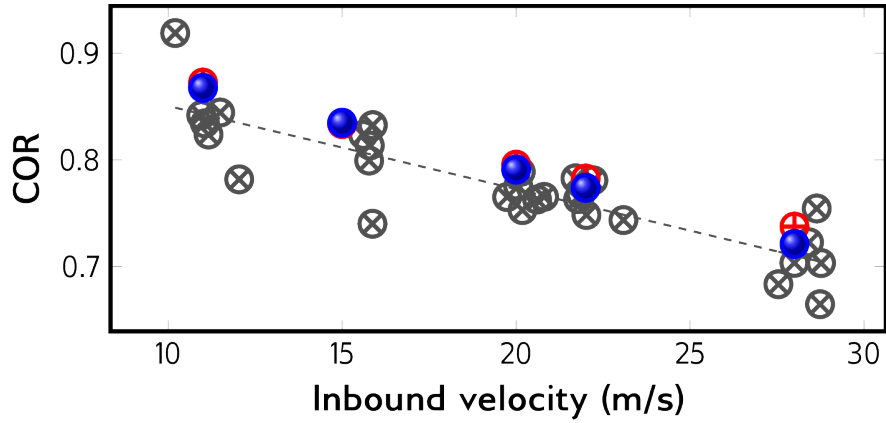

Fig. 2.6: Comparison between the predicted coefficient of restitution of the developed FE model, the advanced model of Price et al. [4] and dynamic impact test.
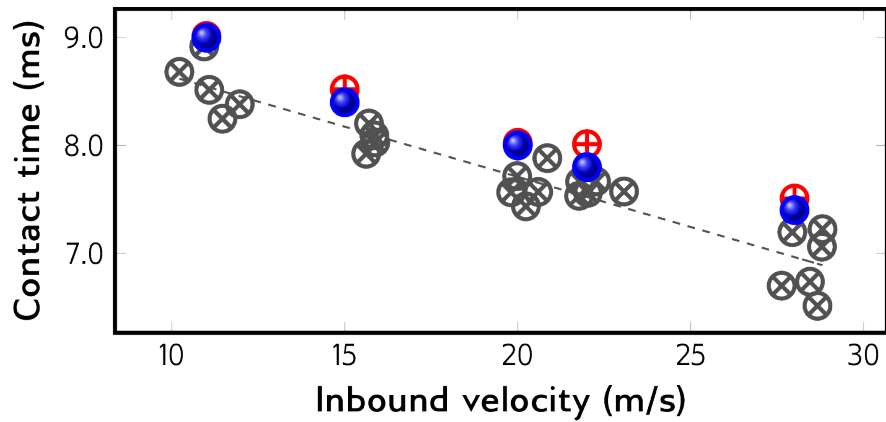

Fig. 2.7: Comparison between the predicted contact time of the developed FE model, the advanced model of Price et al. [4] and dynamic impact test.
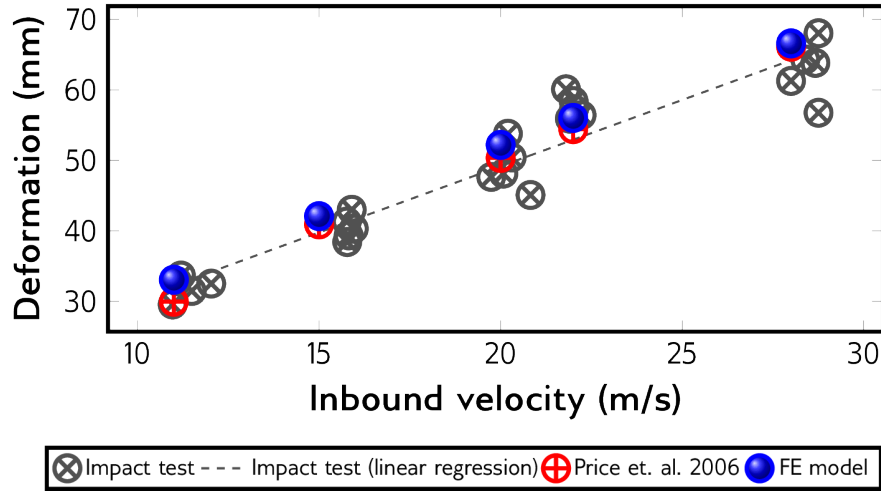
Fig. 2.8: Comparison between the predicted maximum longitudinal deformation of the developed FE model, the advanced model of Price et al. [4] and dynamic impact test.

2.7 Discussion

The values predicted for all parameters were in good agreement with those obtained by Price et al. [4]. The comparison between both models for coefficient of restitution and contact time exhibits almost no discrepancy with only 3% of the maximum difference. The longitudinal deformation shows a 10% difference for the inbound velocity of 11 m/s, but the discrepancies between both models reduce to a maximum of 4% as the inbound velocity increases.

   Comparing with the published dynamic impact test data, it is apparent that the developed FE model predicted slightly higher COR and contact time compared to the linear regression line of the impact test data. Nonetheless, the differences are minimal and almost negligible. The prediction of the maximum longitudinal deformation, on the other hand, was very close to the linear regression line of the impact test data. This shows the capability of the developed FE model in predicting the ball responses during an impact.

2.8 Summary

An FE model of soccer ball was developed. It was modelled using a composite shell comprising of two distinctive layers. The material properties of each layer were

obtained from published tensile test data. The pressurisation of the ball was done by creating a reference node at the centre of gravity of the ball, which was coupled to the inner surface of the ball. Dynamic impact tests were simulated by impacting the ball onto a rigid wall. The predicted coefficient of restitution, contact time and longitudinal deformation of the ball were compared with those of a more advanced soccer ball model developed by Price et al. [4]. A good agreement was achieved, which shows that the soccer ball model developed is capable of producing accurate results. The ball model will be used in simulating soccer heading.

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Chapter 3
Human Head Finite Element Model

Abstract This chapter discusses the development of a simplified human head finite element model. It comprises of four components; the skull, facial bones, cerebrospinal fluid (CSF), and the brain. The material properties of each component are adopted from various literature.

3.1 Introduction

Experimental works have been performed to extract information on the various processes involved during head impacts on cadavers due to their mass distribution and the physical geometry of a living human [1]. Cadavers are also preferred as it is not possible to obtain essential readings such as the head acceleration response, maximum von Mises stress in the brain and the intracranial pressure through non-intrusive means on living humans. Amongst the notable relevant experimental works on human cadavers were performed by Nahum and Smith [2], Nahum et al. [3], Trosseille et al. [4], Hardy [5].

Analytical modelling by means of representing the brain or head as a mass-spring-damper system was also explored by researchers in investigating blunt TBI in sports [6, 7]. Nonetheless, such models restricts the analysis with respect to only the linear as well as angular kinematics response of the brain without providing further insights as well as visualisation of other essential parameters in understanding the mechanics of blunt TBI. Therefore, researchers have opted for FE modelling, as such continuum model provides and predicts better physical as well as mechanical response of blunt impacts on brain.

The simplest human head 2D model was developed by Hardy and Marcal [8] that took the shape of a spherical or ellipsoidal shell as the skull by defining it as an elastic material. Another 2D model assumes the brain as an elastic material which is connected to a skull that is represented as a closed rigid medium [9]. Further improvements took into consideration the inclusion of the scalp and intracranial contents [10], apart from the development of more anthropometric and anatomical

based models with different type of material properties [11]. With the advent of advanced computational technology, complex 3D human head models have been developed [12, 13, 14, 15, 1]. The employment of computed tomography (CT) and magnetic resonance imaging (MRI) has paved the way towards the development of biofidelic FE model [13].

However, it is worth noting that such sophisticated models require high computational cost and whether such models are necessary in reproducing good approximation of the experimental results remain to be seen. Therefore, one of the goals of this study is to develop a relatively simple 3D head model that replicates well the brain response subjected to blunt impact as per Nahum et al. [3] experimental work. The developed model is aimed to be utilised in analysing the impact on the brain caused by heading soccer ball.

3.2 Model Geometry and Material Properties

A human skull geometry was adopted from a previous work [16]. It consists of two main parts: the cranium and the facial bones. The cranium comprises of eight bones that are separated by sutures. These sutures were removed and all eight bones were merged to form a unified human skull. To further reduce the complexity of the model, the lower jaw and the teeth were also removed. Figure 3.1 shows the components of the head FE model.

The cranium and facial bones were defined as linear elastic with an elastic modulus of 6.5 GPa and Poisson's ratio of 0.2 [17, 18]. The density was defined as 3,312 kg/m^3 for the cranium and 8,000 kg/m^3 for the facial bones; these values correspond to the mass of the 50th percentile male head. Underneath the cranium, a 1.3 mm cerebrospinal fluid (CSF) was modelled. The CSF separates the skull and the brain, whilst protecting the brain from various closed head injuries [17] by acting as a natural shock absorber [19]. It was modelled as a nearly incompressible elastic material with an elastic modulus of 150 kPa, Poisson's ratio of 0.49886 and density of 1,004 kg/m^3 [18].

The brain was modelled with solid elements, which fills the area beneath the CSF. It was defined as viscoelastic with a linear elastic material model. The elastic modulus, E of the brain was computed from the bulk modulus, K and shear modulus, G from Ruan et al. [20] using the following equation:

$$E = \frac{9KG}{3K + G} \tag{3.1}$$

which gives a value of 5.04 MPa. The Poisson's ratio and density of the brain was defined as 0.4996 and 1,040 kg/m^3, respectively. The shear characteristics of the viscoelastic behaviour of the brain were expressed by the following equation:

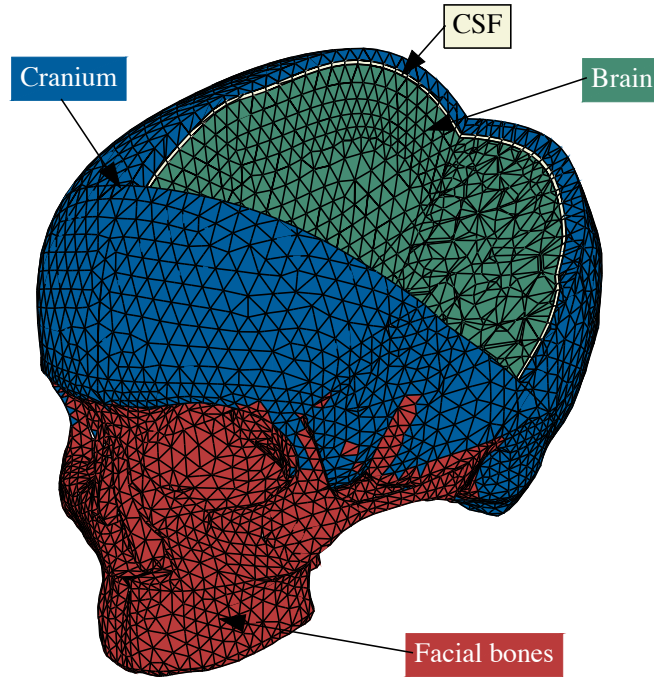$$G_t = G_\infty + (G_0 - G_\infty)e^{-\beta t} \tag{3.2}$$

Fig. 3.1: View cut of the FE head model showing all four components.

where G_0 is the short-term shear modulus with a value of 0.528 MPa, G_∞ is the long-term shear modulus with a value of 0.168 MPa, β is the decay factor that was defined as 35 s^{-1} and t is time, expressed in seconds[21]. These characteristics were implemented in ABAQUS 6.13 as the time domain viscoelastic material model that is given by a Prony series expansion of the following dimensionless relaxation modulus Dassault Systèmes [22]:

$$g_R(t) = 1 - \bar{g}^P(1 - e^{-t/\tau^G})$$

(3.3)

where $\bar{g}^P = (G_0 - G_\infty)/G_0$ and $\tau^G = 1/\beta$ [23]. Altogether the head FE model consists of four sections: the cranium, facial bones, CSF and brain with a total mass of 4.54 kg, which corresponds to the Hybrid III 50th percentile male human head dummy. Table 3.1 summarises the elastic material properties of the head FE model. Since the skull fracture is not of interest, the cranium and facial bones were defined as rigid bodies. All nodes were tied to a reference point at the centre of mass of the head. This method was found to reduce the computation time by more than 80%. It was reported that for a very short duration impact, the neck has minimal influence on the brain responses [20, 24]. A typical duration of impact between 15 to 33 ms

Table 3.1: Elastic material properties.

Component	Density (kg/m^3)	Elastic modulus (MPa)	Poisson's ratio
Skull	3,312	6,500	0.2
Facial bones	8,000	6,500	0.2
Cerebrospinal fluid	1,004	0.15	0.49886
Brain	1,040	5.04	0.4996

in a soccer heading manoeuvre [25, 26, 27, 28] is considered very short. Therefore, a free boundary condition (no constraint) at the head-neck joint was defined.

3.3 Model Validation

Most of the head FE models were validated against the experimental data of Nahum et al. [3]. In their experiments, the frontal skull of a seated stationary human cadaver was impacted by a rigid mass travelling at a constant velocity. The head was rotated forward such that the Frankfort plane of the head was inclined by 45° to the horizontal. The parameters measured in their experiments were the intracranial pressures in addition to the impact force and head acceleration. Five intracranial pressures were measured using pressure transducers placed at several locations: in the frontal bone at the impact contact area, immediately posterior and superior to the coronal and squamosal sutures respectively in the parietal bone, inferior to the lambdoidal suture in the occipital bone and in the occipital bone at the posterior fossa. These transducers were placed within the skull in such way that they would not injure the brain upon impact.

To validate the head FE model, the experiment 37 conducted by Nahum et al. [3] was simulated. Instead of modelling the rigid mass impacting the frontal skull, the impact force from Nahum's experiment was extracted as shown in Figure 3.2; this is applied in the form of pressure on the frontal bone of the skull within an area of 1,630 mm^2. Intracranial pressures were measured at the locations shown in Figure 3.3, as well as the linear acceleration of the head's centre of mass. These locations are the approximate locations of which Nahum et al. [3] have placed the pressure transducers in the experiment. The intracranial pressure is the average of the principal stresses at the selected elements [20].

3.4 Skull-brain interface

The skull-brain interface condition has been modelled using several methods such as tied nodes, sliding interface and the inclusion of solid CSF. The head model in this study incorporates a solid CSF between the skull and the brain. Tie constraint

Fig. 3.2 The impact force
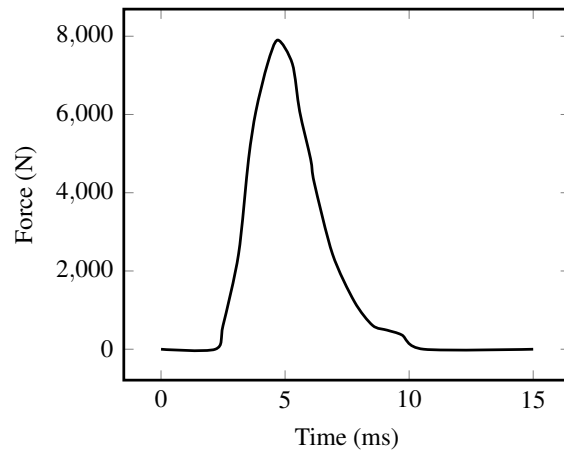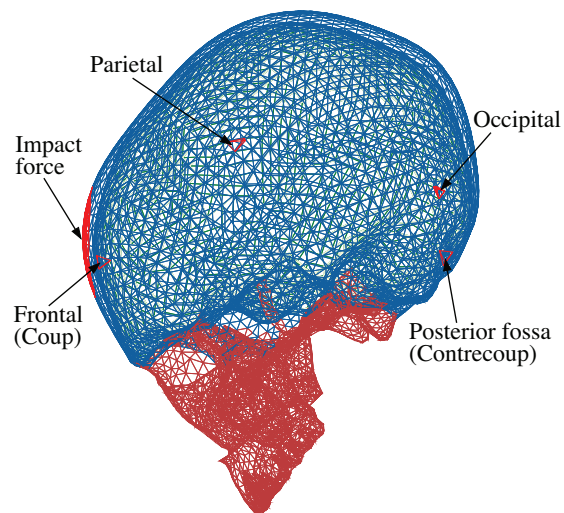extracted from Nahum et al.
[3]

Fig. 3.3 The locations of
the applied impact force
and the intracranial pressure
measurement

properties was defined between the inner surface of the cranium and the outer surface of the CSF, as well as the inner surface of the CSF and the outer surface of the brain. It was reported that the tie constraint interface provides the best agreement with experimental data [29].

Figure $3.4 - 3.8$ show the comparison of intracranial pressures and head acceleration between Nahum's experimental data and those predicted by the FE model. It is evident that the tie constraint interface provides the best agreement with the experimental data. Nonetheless, the head acceleration was underestimated by 17%.

Fig. 3.4 Predicted frontal
(coup) pressure compared to
that of Nahum et al. [3].
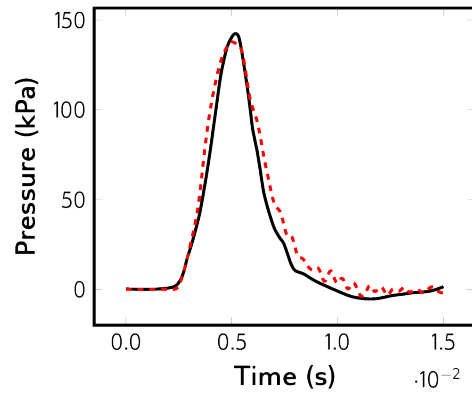
Fig. 3.5 Predicted posterior
fossa (contrecoup) pressure
compared to that of Nahum
et al. [3].
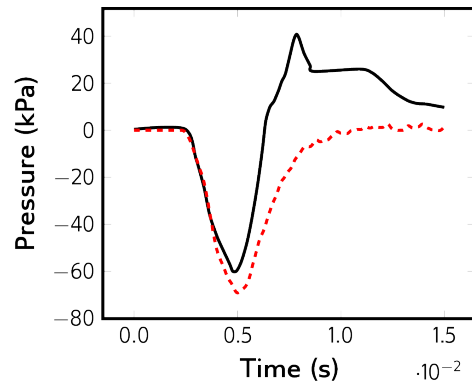
Fig. 3.6 Predicted parietal
pressure compared to that of
Nahum et al. [3].
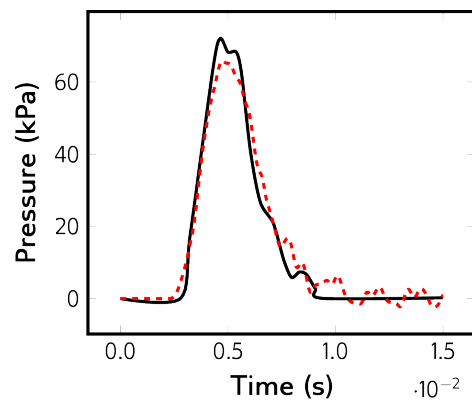
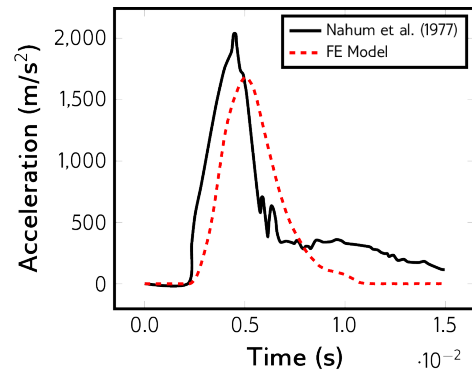**Fig. 3.7** Predicted occipital pressure compared to that of Nahum et al. [3].
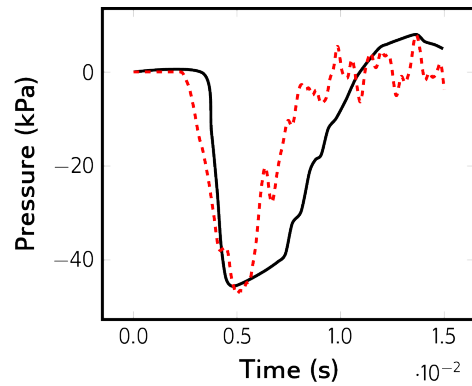
Fig. 3.8 Predicted head acceleration compared to that of Nahum et al. [3].

This might be attributed to the undocumented mass of the cadaver's head used in Nahum's experiment. Since the head acceleration is defined by the equation of motion, the mass of the head has an influence on the head acceleration. The mass of Nahum's cadaver might be smaller than the 50th percentile male head mass, thus resulting in slightly larger head acceleration than that obtained from the simulation.

3.5 Summary

A relatively simple human head FE model was developed. The geometry was adopted from a previous study. The material properties of each component of the head model was obtained from literature. The model was validated against Nahum et al. [3], a cadaveric experimental data that has become a de facto standard in validating a human head FE model. Tie constraint interface between the skull, CSF and

brain was found to provide the best agreement with Nahum et al.'s experimental data. This validated head model will be used in simulating soccer heading. In the next chapter, the FE analysis of soccer heading are presented.

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Chapter 4
Simulation of Soccer Heading Manoeuvre

Abstract In the previous chapters, the FE model of soccer ball and human head were developed and validated against published experimental data. This chapter covers the simulation of a soccer heading manoeuvre, whereby the aforementioned models were assembled and positioned accordingly to replicate an actual soccer heading manoeuvre. The subsequent sections detail the simulation methods, as well as discuss the results obtained.

4.1 Introduction

Ponce et al. [1] and Chen et al. [2] attempted to simulate soccer heading by means of finite element analysis. The former study used a human head FE model, and performed a static analysis by applying a load representing the maximum force exerted by the soccer ball on the forehead of the head model. This method, however, might not be ideal since it neglects the effect of the ball's stiffness, damping and deformation. The latter study, on the other hand, was not documented in detail since it was published as a two-page conference proceedings. Other than these papers, no other study on the FE analysis of soccer heading was found to date. FE analysis provides a lot more flexibility than the experimental work in studying the soccer heading manoeuvre. Therefore, this chapter details the simulation of soccer heading manoeuvre to measure the linear and angular head accelerations throughout the impact. The HIP was then calculated from the acceleration data.

4.2 Soccer Heading Simulation

The soccer ball and human head FE models were positioned accordingly to replicate a heading manoeuvre as shown in Figure 4.1. The FE analysis consists of 4 steps, as follows:

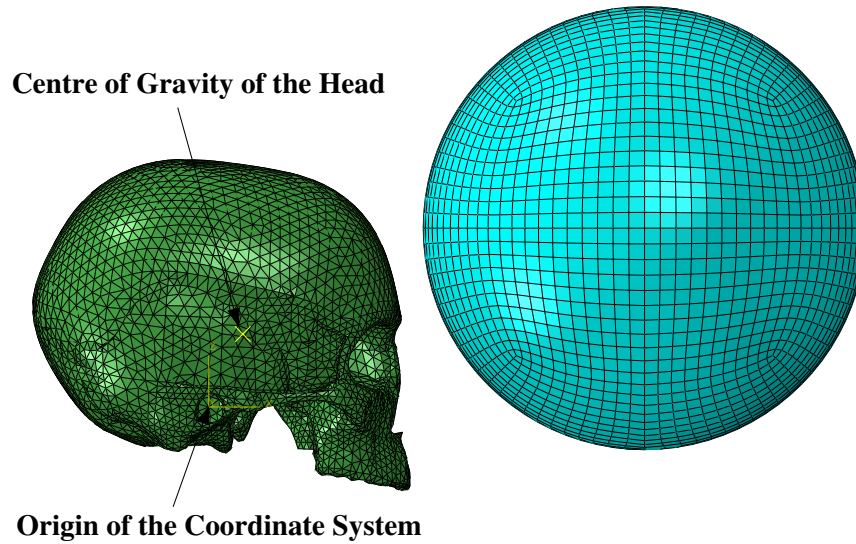**Centre of Gravity of the Head**

Origin of the Coordinate System

Fig. 4.1: Assembly of the FE models to replicate the experiment.

1. Pressurisation of the ball to 60 kPa, which corresponds to the ball pressure in the experiment.
2. A 1 ms 'hold' step to allow the pressurised ball to become stable.
3. Definition of the desired ball velocity, but the velocity definition was turned off at the end of the step to allow for impact.
4. The ball impacted the head and rebounded.

Five FE analyses were performed with the ball inbound velocities of 4, 6, 8, 10 and 12 m/s. Figure 4.2 shows the sequence of the impact from the FE simulation for the ball velocity of 8 m/s. It illustrates the impact as the ball came in contact with the head until it left 14 ms afterwards. It is apparent that the FE analysis produced comparable impact characteristics to the actual soccer heading manoeuvre based on the observation of the ball contact time and its deformation.
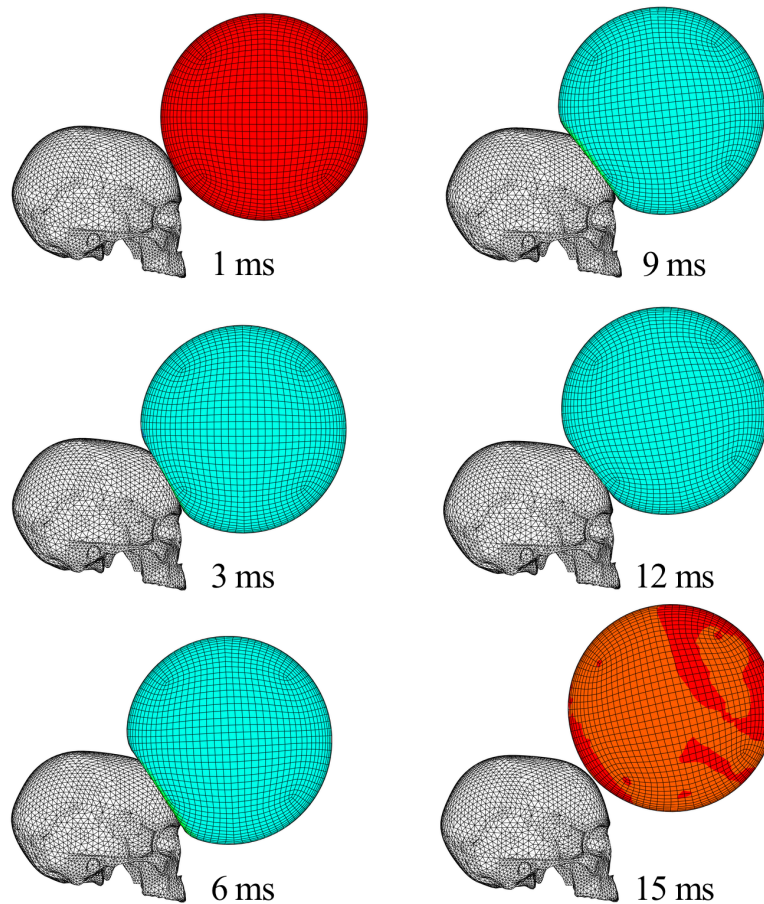
Fig. 4.2: The sequence of simulated soccer heading manoeuvre.

4.3 Validation against Published Experimental Works

This section covers the validation of the FEA against published experimental works. As previously mentioned, several studies were conducted to measure the linear and angular head acceleration during soccer heading [3, 4, 5]. Figure 4.3 and 4.4 depicts the peak linear and angular head accelerations obtained from these studies and those predicted by the FEA.   The predicted accelerations are in a good agreement with those of literature with the linear regression lines of both data are very close. The

Fig. 4.3 Comparison of the predicted and published linear head accelerations due to soccer heading.
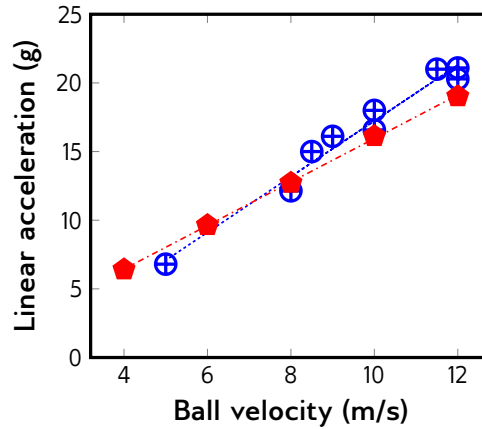
Fig. 4.4 Comparison of the predicted and published angular head accelerations due to soccer heading.
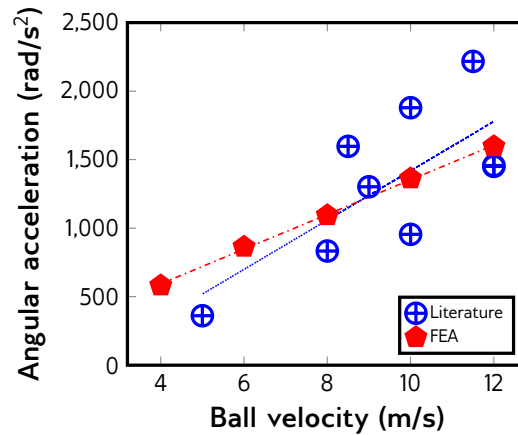
detail comparison between the results obtained by the FEA and those of literature is presented in this section.

Amongst the earliest experiment conducted to measure the head accelerations induced by soccer heading was conducted by Naunheim et al. [3]. The experiment was performed on human subjects wearing a headpiece instrumented with four accelerometers. The accelerations recorded were transformed to the centre of mass of the head. A soccer ball inflated to 55.3 kPa was projected at the subjects at inbound velocities of 9 and 12 m/s. The subjects were asked to head the ball back towards the launching machine. Figure 4.5 shows the comparison between the peak linear and angular head accelerations measured by Naunheim et al. and those predicted by the FEA. Naunheim et al. did not mention about the head velocities in the ex-
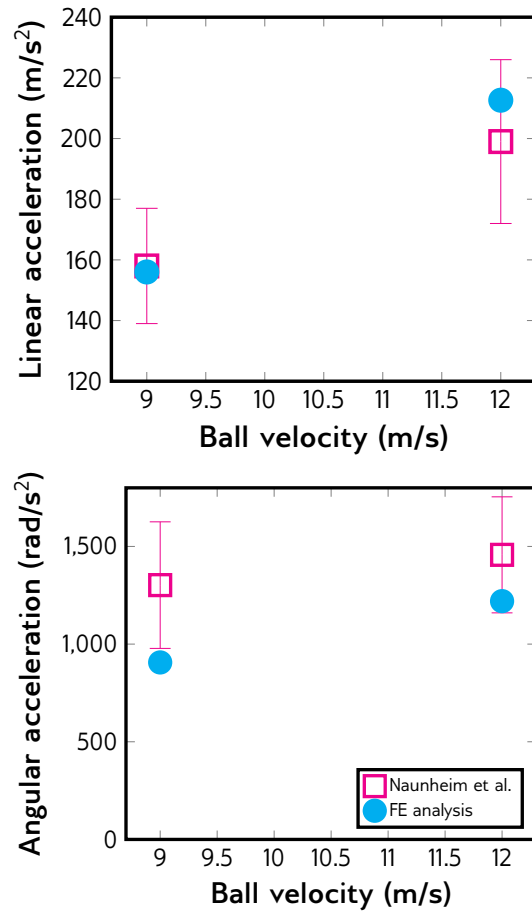
Fig. 4.5: Predicted peak linear head accelerations and those of Naunheim et al. [3].

periment, nor the impact location. Thus, one could only conjecture the location of the impact, and this has a significant effect on the prediction. Taking into account the standard deviation of the experimental data, the closing speed and the unknown impact location, it is evident that the FEA produced comparable results.

Besides human subjects, soccer heading experiments were also performed on Hybrid III crash test dummy headform. The headform has a mass of 4.54 kg and it is equipped with 9 accelerometers. Hanlon and Bir [4] conducted a series of soccer ball impact tests on the headform at ball velocities of 8, 10 and 12 m/s. The headform was subjected to the soccer ball impact at the forehead, left temple and right side. It was also fitted with a headband equipped with a wireless head acceleration measurement system known as the Head Impact Telemetry System (HITS). This system provides a real-time measurement of head accelerations. The accel-

erations measured by the Hybrid III headform and those measured by HITS were compared to assess its accuracy. Their experiments were simulated and the results compared. Figure 4.6 shows the linear acceleration curve for 12 m/s ball impact on the forehead, left temple and right side. The FEA slightly overestimated the linear accelerations in all cases, however the acceleration profiles predicted were in a better agreement with those of Hybrid III headform compared to the HITS. The exact impact locations as well as the head angle during their tests were not documented, which might attribute to the discrepancies between the experimental results and simulation. Nonetheless, the differences are considered minimal, and the FEA was able to estimate the linear accelerations of each impact locations satisfactorily. The duration of the acceleration was also comparable to that of the Hybrid III and HITS in all cases.

In addition to the linear accelerations, the angular accelerations were also compared. However, no angular acceleration curves were made available by the authors. Thus, only peak linear and angular accelerations for every impact velocity of the respective impact locations were compared as depicted in Figure 4.7a − 4.7f. Owing to the lack of information on the exact impact locations in the experiments, slight discrepancies were observed for the linear accelerations between the FEA and the Hybrid III headform for all impact locations. The FEA overestimated the linear accelerations in all cases; this is attributed to the absence of the neck in the model. The flexible rubber neck of the Hybrid III headform might have absorbed some of the impact forces, thus reducing the resulting linear accelerations slightly. Nonetheless, in some cases, the FEA results were in a good agreement with the HITS.

The agreement between the FEA results and those of Hybrid III headform was better for the angular accelerations. The FEA results were better than the HITS in almost all cases. The head accelerations were measured directly at the centre of mass in the case of the Hybrid III and the FEA, whilst in the HITS, the accelerations were measured at several locations and they were transformed to the centre of mass. This attributes to the better agreement between the Hybrid III and FEA as compared to the HITS. The angular accelerations produced by the FEA fall within the standard deviation of those recorded by the Hybrid III. Figure 4.5−4.7 demonstrate the capability of the FE models in producing accurate results for soccer heading analysis. It was observed that the model assembly (i.e. the head angle and ball impact location) has a significant influence on the predicted head accelerations. Even a small difference leads to a discrepancy between the predicted and experimental data. This suggests that the FEA is superior than experimental work in studying soccer heading to prevent from these inconsistencies. The subsequent section discusses a parametric FE analysis of soccer heading to investigate the influence of several parameters on the linear and angular head accelerations.
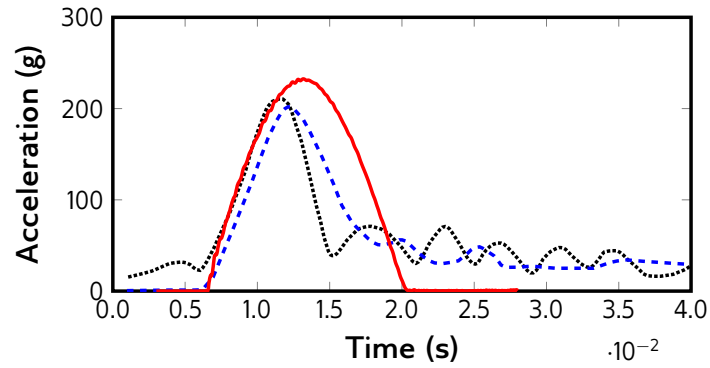
(a) Forehead
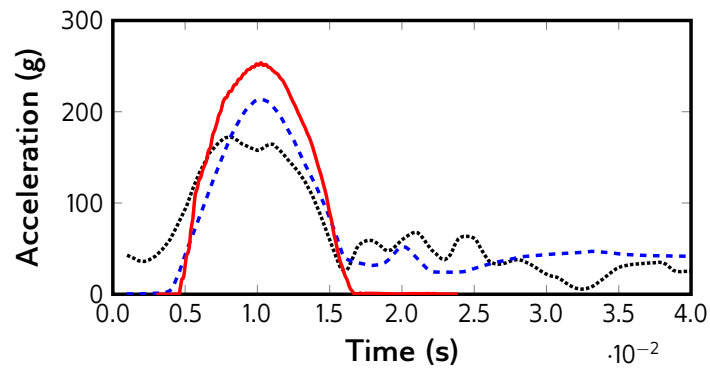
(b) Left temple
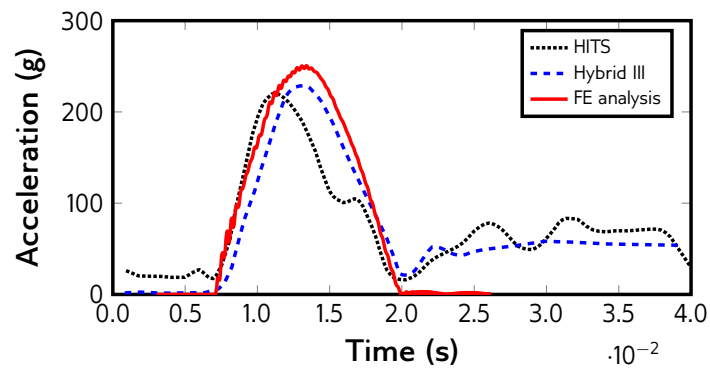
(c) Right side

Fig. 4.6: Predicted linear head acceleration curves and those of Hybrid III headform, HITS headgear for 12 m/s soccer heading
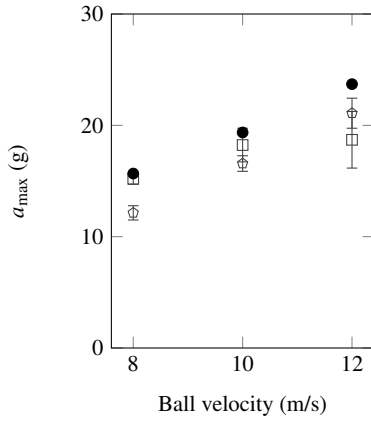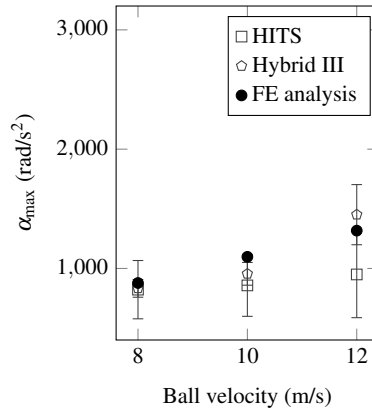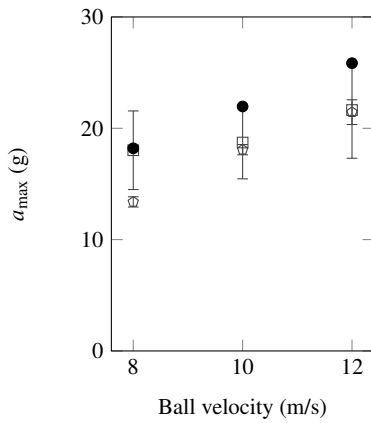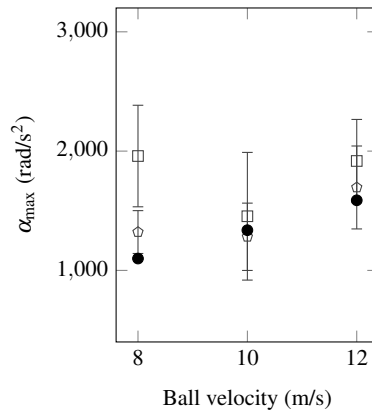
(a) a_{max} - forehead impact

(b) α_{max} - forehead impact

(c) a_{max} - left temple impact

(d) α_{max} - left temple impact
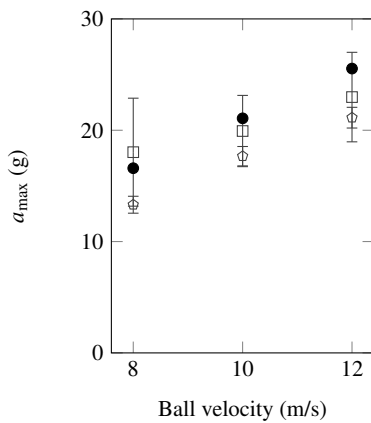
(e) a_{max} - right side impact
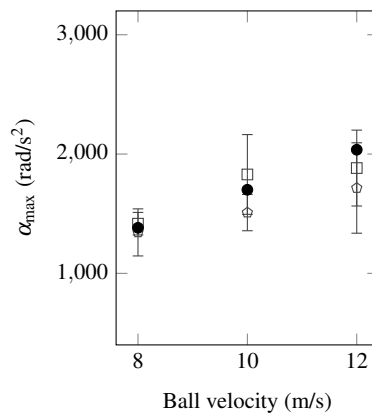
(f) α_{max} - right side impact

Fig. 4.7: Predicted peak linear (a_{max}) and angular accelerations (α_{max}) plotted alongside those measured by the Hybrid III headform, HITS [4] for the impacts at the forehead, left temple and right side.

4.4 Summary

The developed soccer ball and human head FE models were assembled to replicate soccer heading manoeuvre. Soccer heading simulations were performed for a range of inbound velocities of 4 to 12 m/s. The predicted linear and angular head accelerations were compared with those of literature, which demonstrates an acceptable agreement. This suggests that the FE models developed were capable of predicting the linear and angular accelerations of the head due to soccer heading manoeuvre.

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Chapter 5
Analysis of Protective Headgear

Abstract In the previous chapter, soccer heading simulations were performed and validated against literature. It is evident that the FEA predictions of the linear and angular head accelerations were in a good agreement with the published experimental data of soccer heading. In this chapter, similar simulations were conducted, but with the inclusion of a headgear modelled on the forehead of the head model. The subsequent sections describe the methods and discuss the results obtained from the aforementioned FE simulations.

## 5.1 FE Modelling of the Soccer Headgear

Fig. 5.1 Full90 Premier commercial soccer headgear.

A commercial soccer headgear (Full90 Premier as depicted in Figure 5.1) was utilised in the analyses of this chapter. The headgear is made of high-density polyethylene foam. The material properties of the foam were adopted from Lehner

[1]. He conducted a compression test on the foam and the compressive stress-strain curve obtained is shown in Figure 5.2.

Fig. 5.2 Stress-strain curve of the foam used in the Full90 headgear obtained from a compression test [1].
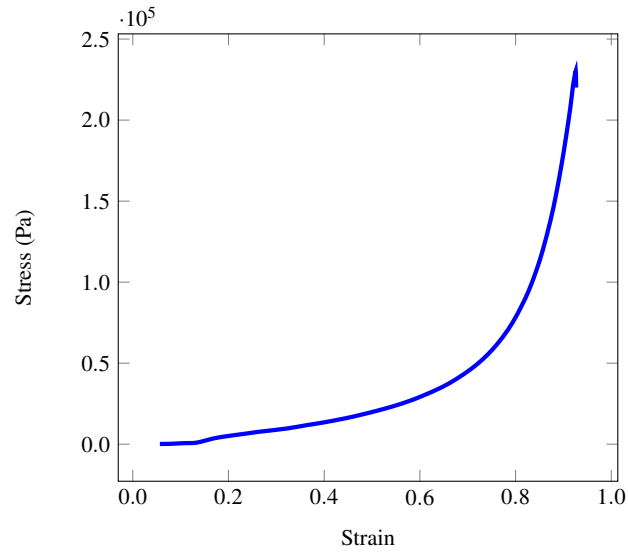
Solid meshes were created from the surface of the frontal skull to represent the headgear as depicted in Figure 5.3. The compressive stress-strain curve from

Fig. 5.3 Headgear modelled on the frontal skull. (*Blue elements represent the headgear*)
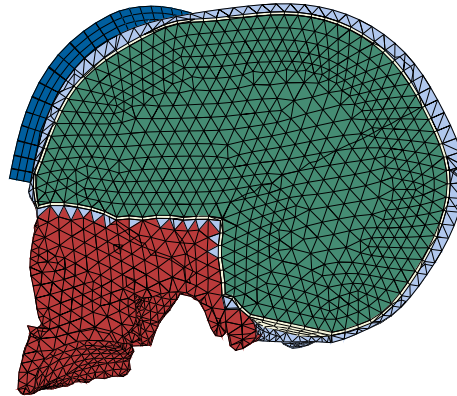
Lehner [1] was extracted and was used to define the properties of the Full90 headgear. The hyperfoam material model was employed in Abaqus/CAE. A high-density elastomeric foam usually has a density between 500 to 1,000 kg/m^3 [2]. Thus, the

density of the headgear was defined as 900 kg/m^3. Assuming zero lateral expansion, the Poisson's ratio was defined as 0.02, close to zero since a zero Poisson's ratio resulted in an error during the computation. Linear and angular head velocities and accelerations were requested from the simulations. Five FE simulations were performed, with the ball inbound velocities of 4, 6, 8, 10 and 12 m/s.

5.2 Results and Discussions

Fig. 5.4 Predictions of linear head accelerations with and without the presence of the headgear.
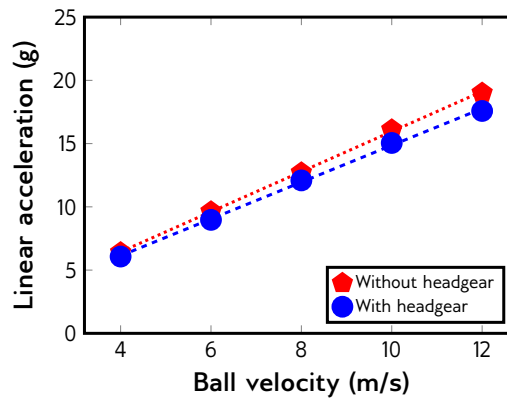

Fig. 5.5 Predictions of angular head accelerations with and without the presence of the headgear.
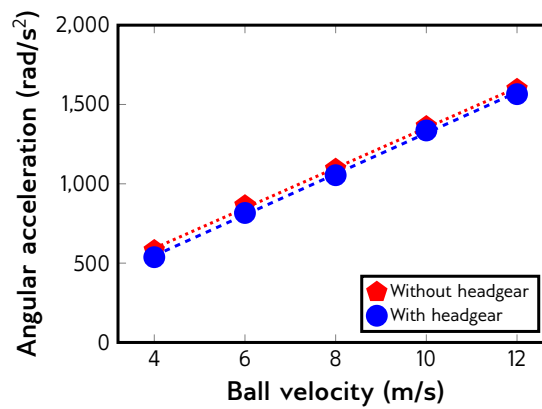
Figure 5.4 and 5.5 shows the peak linear and angular accelerations of the simulations without wearing the headgear as depicted in the previous chapter, plotted against those of with wearing the headgear obtained from the aforesaid simulations. It is evident that the two lines (red and blue) in both figures are very close to each other, with the maximum difference of not more than 10%. This suggests that the use of the Full90 Premier headgear when performing soccer heading manoeuvre does not have a significant influence on the resulting linear and angular head accelerations. The results obtained are congruent with the findings of previous studies [3, 4, 5, 6].
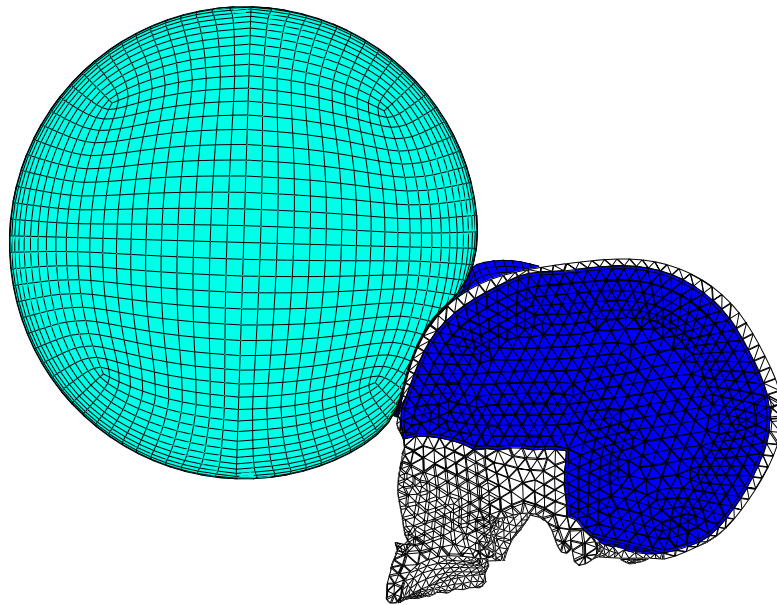

Fig. 5.6: Headgear was fully deformed during the soccer ball impact.

The FE analysis also allows for a closer observation of the deformation of the headgear during impact. This deformation is very difficult to be monitored in the experiment without an expensive high-speed video equipment. It has been mentioned that in order for a headgear to be effective in mitigating the impact during a soccer heading manoeuvre, it should not deform fully. From the FE analysis, it is apparent that the Full90 headgear was fully compressed by the soccer ball even at the lowest tested ball velocity of 4 m/s. Figure 5.6 illustrates the deformation of the headgear during the impact. The fact that the Full90 headgear deforms fully when it is impacted by the soccer ball even at a low ball velocity suggests that it is incapable

of protecting the head from sustaining head injuries due to soccer heading. This was also reported by Naunheim et al. [7].

5.3 Summary

This chapter investigates the influence of a protective headgear on the linear and angular head accelerations and the risk of sustaining head injury due to soccer heading. The results suggest that the headgear is incapable of mitigating the head acceleration and thus the risk of head injury due to soccer heading. FE analysis demonstrated that the headgear was fully deformed upon impact with the soccer ball, thus making it inefficient in reducing the head response due to the ball impact. It is evident that the FE analysis was able to replicate the soccer heading manoeuvre satisfactorily, hence can be used as a tool in evaluating and designing a headgear for soccer players that could reduce the impact from the soccer ball. Future works will look into the design of the headgear, such as the shape of the foam, the composite of foams, and so on.

References

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