An *in-silico* study of the effect of non-linear skin dynamics on skin-mounted accelerometer inference of skull motion

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Highlights:

- An accurate, nonlinear, second-order skull-skin-sensor system model was developed
- A linear comparator assessed the effects of soft tissue artefact in impact response
- The non-linear model drastically overestimated head acceleration during impact
- Thus, skin mounted sensors are likely to overestimate acceleration of heavy impacts

In Brief:

The system of an accelerometric sensor mounted to the skin is often used to assess the accelerations experienced by athletes during impacts. By developing a non-linear model of the skull-skin-sensor model, the importance of accounting for the non-linear nature of skin while taking telemetry measurements of head acceleration during impact is demonstrated.

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Abstract: Accurate and precise analysis of head impact telemetry data is important for development of biomechanical models and methodologies to decrease the risk of traumatic brain injury. Systematic review suggests that much existing data lacks verification. Soft tissue artefact is a common problem that is not frequently addressed. This paper outlines a method of modelling the coupled, non-linear, skull-skin-sensor system. The model is based on a first order underdamped spring mass damper system that incorporates non-linear values to account for the complex dynamic nature of skin. MATLAB was used to simulate the estimated movement of a sensor mounted to the skin relative to measurements collected via a mouthguard sensor. The non-linear elastic and damping models were developed from descriptions in literature. The model assumed a sensor of 8g, mounted behind the ear. Results were compared to a typical linear system. In small impacts, the linear and non-linear models provided similar accelerations to the skull. However, in large impacts, the acceleration of the sensor was estimated to be 158% greater than the skull acceleration when modelled non-linearly, while a linear model showed only a 0.7% increase. This implies that for small impacts, the nonlinearity of skin-skull dynamics is not an important characteristic for modelling. However, in large impacts, the non-linearity of the skin-skull dynamic can lead to drastic over-estimates of skull acceleration when using skin mounted accelerometers.

Key words: Soft tissue artefact, skin modelling, head impact, skin-mounted sensor, accelerometer.

INTRODUCTION

Traumatic brain injuries can lead to serious morbidity and mortality (El Sayed, Mota, Fraternali, & Ortiz, 2008; Goldsmith & Plunkett, 2004; Mark W. Greve, 2009; Meaney, Morrison, & Dale Bass, 2014; Pierce, Smith, Trojanowski, & McIntosh, 1998). The sudden and unexpected nature of most head injuries coupled with the inability to palpitate the brain has led to a paucity of research in traumatic brain injury. However, recent research in high contact sports has led to an emergence of data relating to head impact kinematics and their cognitive effects (Gilchrist & O'Donoghue, 2000; Rowson et al., 2012; Zhang, Yang, & King, 2004). A 2020 systematic review suggested that over two-thirds of the 168 head impact telemetry papers published in the previous decade may be imprecise due to the methods employed to record, verify and process the impact data (Patton et al., 2020). Patton et al. (2020) emphasise concerns over the reliance of many researchers on invalidated filtering algorithms to remove false positive impacts without corresponding video verification.

A common problem in biomechanics is soft tissue artefact (STA), where skin deforms relative to the underlying bone. This relative motion results in discrepancy between skin-mounted sensor acceleration readings and the acceleration of the underlying physical structure (Lucchetti, Cappozzo, Cappello, & Della Croce, 1998; Shultz, Kedgley, & Jenkyn, 2011). Often due to marker inertia (Shultz et al., 2011), the effects of skin deformation can become exaggerated during short-duration, high-magnitude events measured by head impact telemetry systems (Patton, 2016; Wu et al., 2016). Specifically, telemetry systems adhered to the head via the mastoid process have been reported to substantially over-estimate head acceleration on impact due to STA (Wu et al., 2016). Significant measurement error has also been observed for helmet- and headband-based telemetry systems due to insufficient sensor-skull system coupling, producing excess sensor movement relative to the skull (Cummiskey et al., 2017) (Jadischke, Viano, Dau, King, & McCarthy, 2013). Consequently, there is ongoing reporting of unrealistic head impact telemetry data (Jadischke et al., 2013; D. King, P. Hume, C. Gissane, & T. Clark, 2017a; Patton, 2016; Rowson et al., 2012; Wu et al., 2016).

Accurate reporting of head impact telemetry data is crucial to enable the development of valid injury severity criteria and finite element brain modelling of potentially injurious events (Post et al., 2015). More accurate data could provide a basis for the development of training methodologies and suggest changes to improve athlete brain health and wellbeing. More recent evolutions of head impact telemetry systems include inertial measurement unit (IMU) sensors embedded in mouthguards to improve coupling of skull-sensor systems (A. Bartsch, Samorezov, Benzel, Miele, & Brett, 2014; A. J. Bartsch et al., 2019; Greybe et al., 2020; Wu et al., 2016; Wu, Zarnescu, Nangia, Cam, & Camarillo, 2014). This research presents a model of the dynamics between the skull and an accelerometer placed on the skin behind the ear. The model of the coupled skull and skin/sensor unit system will be used to investigate how similar the motion of skin mounted sensors is to the measurements produced by mouthguard sensors that remain more closely coupled to the head during impact or acceleration.

METHOD

Multiple models of skin kinematics and soft tissue artefact have been developed (Joodaki and Panzer 2018, Lapeer et al. 2009, Sutula 2020, Jor Parker Taberner 2013). However, to ensure the model in this research is based directly on empirical evidence, data presented in Jor, Parker, Taberner, Nash, & Nielsen, (2013; Parker, (2016) will be used as a basis of model development. Parker et al. (2016) conducted a number of experiments perturbing inner forearm skin in a transverse direction to determine multiple stiffness and damping values across a range of displacements. The skin was modelled based on a modified first order underdamped mass spring damper system as defined by Equation 1.

$$\ddot{u}_S = \frac{1}{m}(-k(u_R)\dot{u}_R - c(u_R)u_R) \tag{1}$$

$$u_R = u_S - u_C \tag{1a}$$

Where u_R is the relative displacement between skin (u_S) and the skull (u_C) [m], $\ddot{u}$ is the resultant acceleration [m·s^{-2}], $\dot{u}$ is the velocity [m·s^{-1}], and m the combined mass of a sensor adhered to the skin and effective mass of the moving skin section. In this equation the variables $k(u_R)$ and $c(u_R)$ are the nonlinear functions representing the skin's effective spring constant and damping respectively, which are a non-linear function of displacement. Figure 1 outlines the model visually.

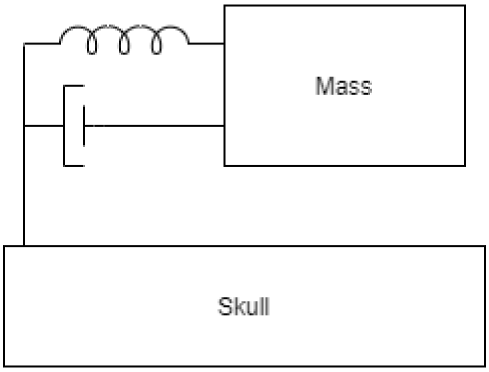

Figure 1: Visual representation of the modelled spring mass damper system.

In this model, the spring and damper contributions were represented by non-linear functions. These were established using MATLAB's least squared curve fitting function (lsqcurvefit) to parameterize data which was collected by Parker (2016). The spring relationship was observed with a least squares analysis of candidate models to have an exponential shape (equation 2).

$$k(u_R) = \Theta_1 e^{\Theta_2 u_R} \tag{2}$$

The damping relationship was determined to be a power function (equation 3).

$$c(u_R) = \Theta_3 u_R^{\Theta_4} + \Theta_5 \tag{3}$$

Substituting Equations 2 and 3 into Equation 1 results in Equation 1b.

$$\ddot{u}_S = \frac{1}{m}\left(-\Theta_1 e^{\Theta_2 |u_R|}\dot{u}_R - \left(\Theta_3 |u_R|^{\Theta_4} + \Theta_5\right)u_R\right) \tag{1b}$$

Where Θ_1 through Θ_4 are equation constants established via the aforementioned curve fitting. It should be noted that the values obtained by Parker (2016) were adjusted due to a suspected error in reported units. When compared to the values reported in Wu et al. (2016), those reported by Parker (2016) were 1000x higher than expected. This disparity was accounted for by correcting the units of spring constant reported in Parker (2016) from N·mm^{-1} to the SI unit of N·m^{-1}.

MATLAB was then used to simulate the skin's response to the acceleration of the skull. Head acceleration data was collected by the head impact research team, (ASTEM, Swansea University, UK), during a collegiate men's rugby game via IMUsinstrumented mouthguards. Institutional ethical approval for the data collection was provided in accordance with the 2013 Helsinki Declaration (SU 2016-059). Before using the data, it was analysed using both frequency decomposition (Shenoi, 2006) and visual verification of impact. A low pass filter (LPF) was implemented via MATLAB's digital filtering toolbox to ensure that high frequency effects did not produce impossible acceleration profiles.

Because the data had been collected via mouthguard sensor, it was multiplied by an affine transform matrix to initially shift the accelerations to the approximate centre of mass of the head, then to the equivalent location of a sensor placed behind the ear. Figure 2 outlines these locations. The simulation was repeated with four datasets corresponding to separate measured impacts throughout the game.

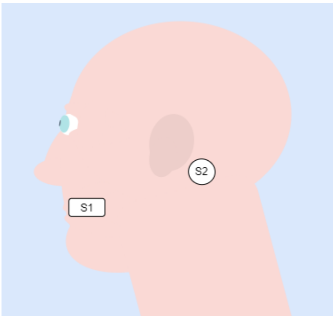

Figure 2: Sagittal view of the relative locations of the mouthguard sensor (S1) and a sensor adhered to the skin behind the ear (S2).

The transformed and filtered data was then used to forward simulate the skin response according to Equation 1b, with the spring and damping constants re-calculated at each time step. The displacement and velocity of the skull was determined using simple time-stepping numerical integration of the acceleration profiles. Initial conditions were set to rest for all displacements and velocities. After simulation, the absolute values of acceleration were considered.

The simulation was then repeated with constant k and c values (Equation 1c). For the linear model, the constants were set based on the values at 1 mm ($k_{linear} = k(0.001), c_{linear} = c(0.001)$). This was used to find the disparity between predicted motion of the skin based on a linear and non-linear model.

$$\ddot{u}_S = \frac{1}{m}(-\theta_1 e^{0.001\theta_2}\dot{u}_R - (\theta_3 0.001^B + \theta_4)u_R) \tag{1c}$$

To investigate the generalisable differences in responses from the linear and nonlinear model, a dual parametric sweep study was performed. This involved running the forward simulation using a single period sine acceleration input. The period of the sine curve was assessed on the range of 8 to 12 ms, and the magnitude range was assessed from 0 to 1000 m·s^{-2}. The peak absolute acceleration of the skin was measured for each model and input condition. This enabled a relationship to be determined between variation of output for the linear and non-linear models' and the magnitude of acceleration at a given period.

RESULTS

The identified non-linear relationship between displacement and effective damping between the skin and skull motion is shown in Figure 3 (left). The fitted non-linear relationship between relative displacement of the skin and skull and Hooke's coefficient is shown in Figure 3 (right). Table 1 provides the established equation parameters according to Equation 2 and 3. Figure 4 compares the movement of the skull and the simulated response of the skin using both the linear and non-linear models for relatively large impact. Figure 5 shows the same comparison for a lower magnitude impact.

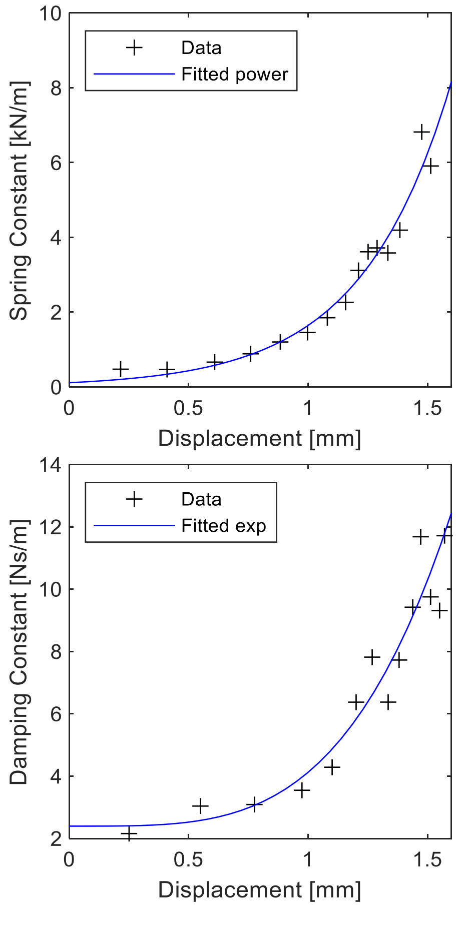

Figure 3: Identified exponential relationship between displacement and effective damping (left) and identified exponential relationship between displacement and effective spring constant of the skin (right).

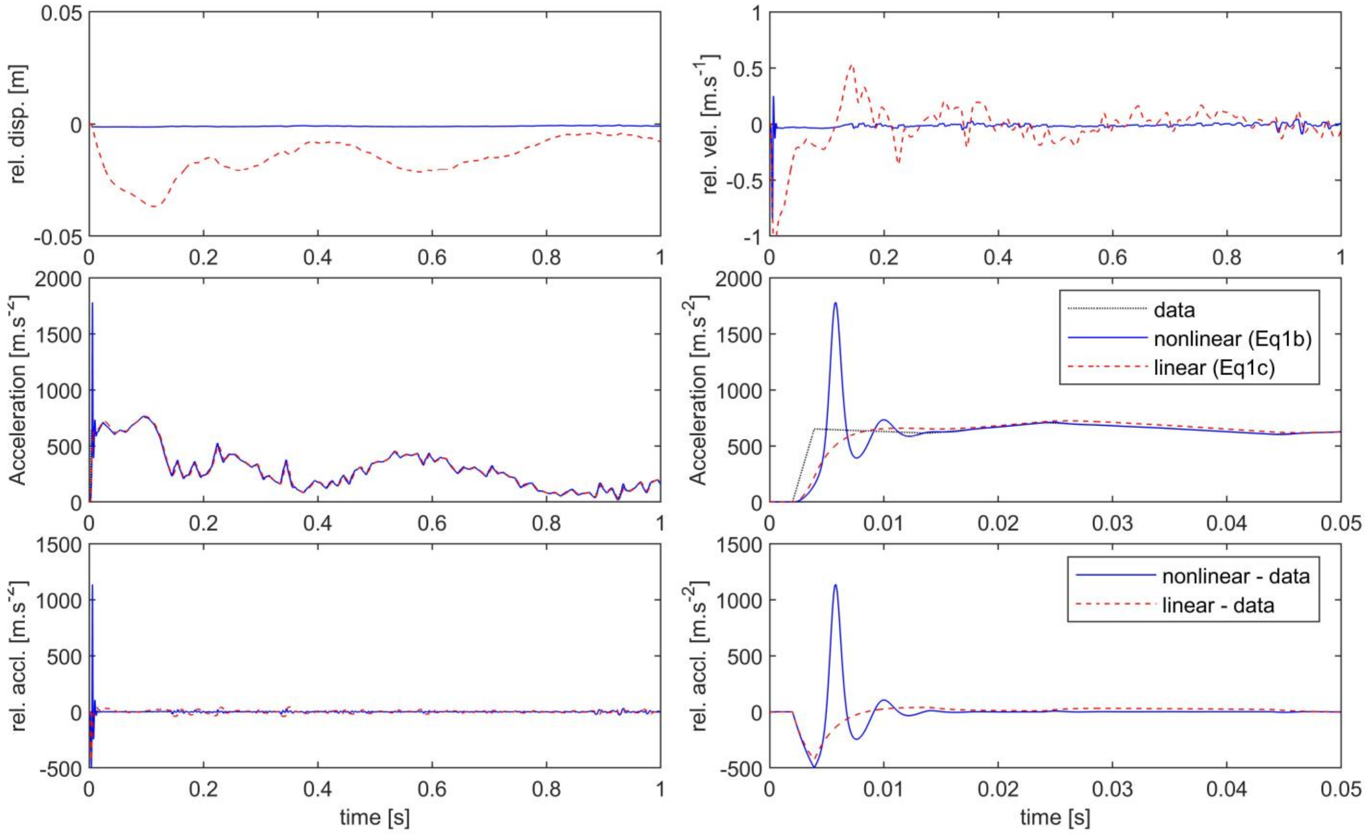

Figure 4: Comparison between the linear and non-linear models and the measured kinematic data for a large magnitude impact. Note the different x-axis scales of the bottom right and middle right plots.

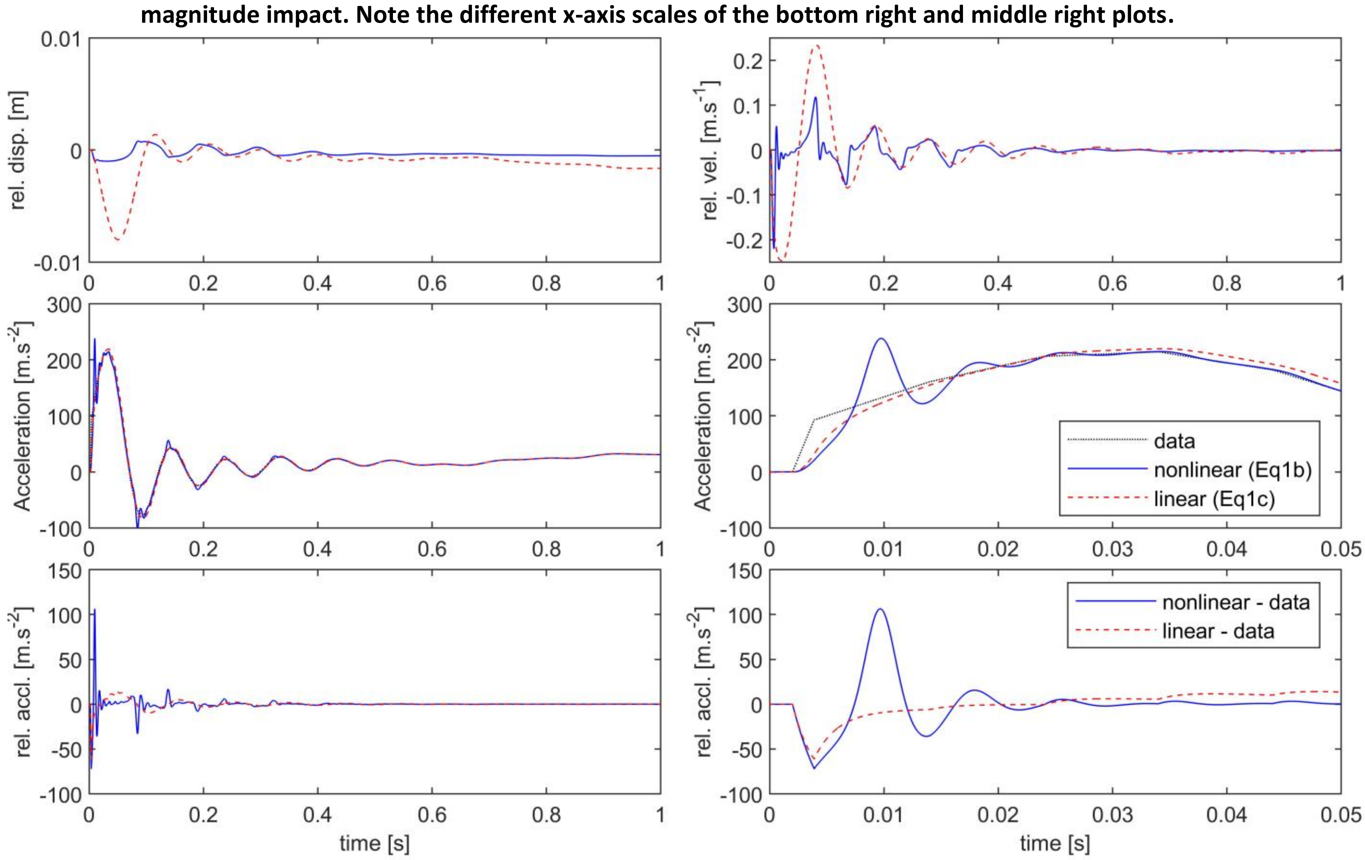

Figure 5: Comparison between the linear and non-linear models and the measured kinematic data for a low magnitude impact.

Table 1: The established parameters for the non-linear spring and damping relationships.

Equation	Parameter	Value
2 (Spring)	θ_1	1.139 x10^5
	θ_2	2.669 x10^3
3 (Damper)	θ_3	3.000 x10^{11}
	θ_4	3.7467
	θ_5	2.396

Table 2 presents a comparison between the peak accelerations of the measured data, the non-linear model and the linear model. Figure 6 shows an example of the difference in response of the linear and non-linear systems over a varying magnitude of acceleration input. The resultant peak accelerations for both the linear and non-linear systems are subsequently plotted according to input signals.

Table 2: Overview of simulation results for four measured impacts. Note the peak recorded acceleration is considered after the affine transformation.

Impact number	Peak recorded acceleration of the skull model input ($\ddot{u}_C$)	Peak simulated acceleration of the skin/sensor model output ($\ddot{u}_S$)	Peak simulated acceleration of the skin/sensor model output ($\ddot{u}_{S,linear}$)
1	735.7 m·s^{-2}	1738.8 m·s^{-2}	740.5 m·s^{-2}
2	360.5 m·s^{-2}	360.6 m·s^{-2}	366.6 m·s^{-2}
3	214.1 m·s^{-2}	238.1 m·s^{-2}	219.8 m·s^{-2}
4	763.3 m·s^{-2}	1780.1 m·s^{-2}	765.6 m·s^{-2}

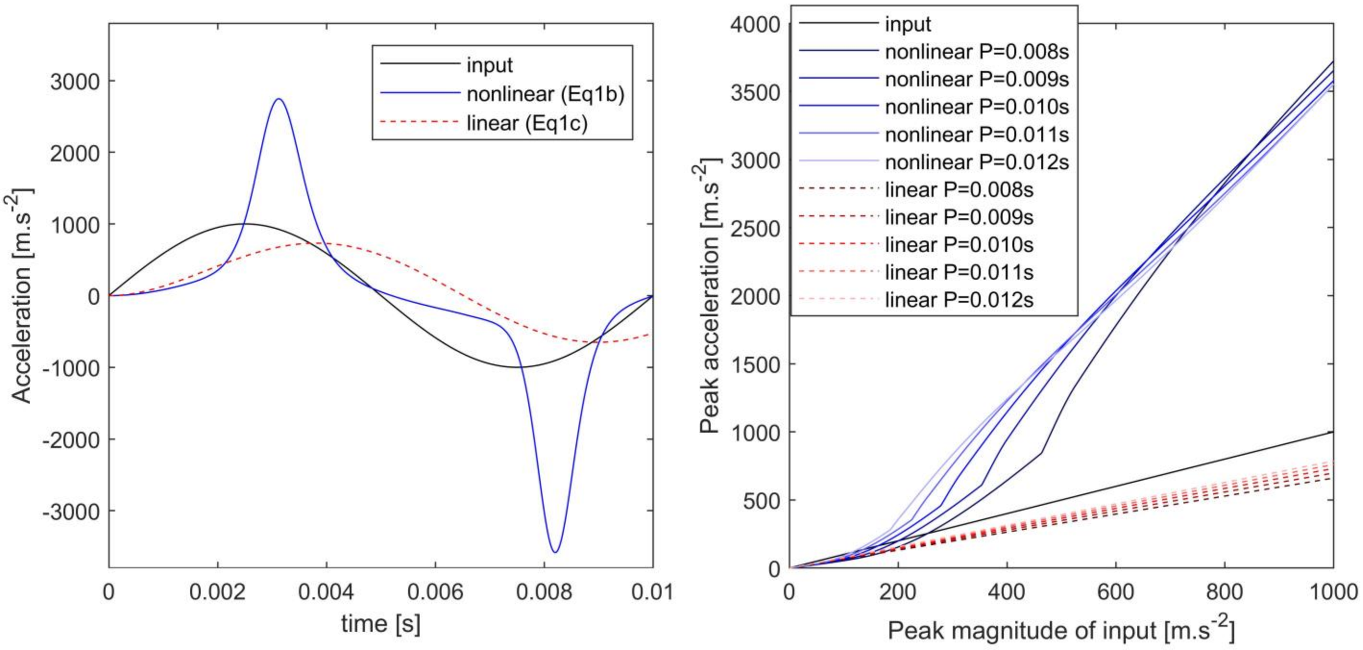

Figure 6: Comparison of the outputs for the linear and non-linear systems to sine wave input (top). Peak simulated accelerations for both the linear and non-linear systems according to the input signal conditions (bottom).

DISCUSSION

Figure 4 shows that a sudden, large acceleration of the head is likely to produce a large increase in acceleration of the skin, compared to the skull. This effect was not present in the linear model. Hence, the increased acceleration observed in this simulation may be due to the non-linear nature of the skin's effective spring and damping properties with respect to relative displacement. If repeated in physical experiments, this implies that in order to get data which closely represents the head's acceleration, a sensor should be attached in such a way that the results do not include STA. This result is in agreement with (Wu et al., 2016). Furthermore, this research has shown that, despite the non-linearity of the skin physiology, for smaller displacements the effects of this non-linearity of the skull-skin-sensor response is negligible. This implies the existence of an inflection point beneath which it may be realistic to model the skin as a linear mass damper system and to assume skull acceleration from skin mounted sensors.

To simplify this analysis, the effects of rotational acceleration on the skin-skull system response were excluded. Rotation was considered in the affine transformation of the acceleration data. If the rotational spring and damping relationship is similarly non-linear, it is possible that this analysis has underestimated the expected accelerations observed on the skin. However, since the accelerometer's centre of mass is very close to the centre of the adhesion area, the rotations of the accelerometer are unlikely to cause significant differences to assumed skull acceleration.

Both the linear and non-linear models resulted in a similar lag time to peak acceleration, when compared to the mouthguard data. This is expected for any second order underdamped spring mass damper system. It can be observed that higher impulse leads to higher expected acceleration of the skin (Figure 6 and Table 2). This highlights the importance of removing any high-frequency interference from measured data. Overlooking this would likely lead to much larger simulated peaks in acceleration of the skin, and possibly even lead to a numerical divergence that must be accounted for in forward simulations. In this case, the filtered acceleration data had sufficient fidelity to allow precise evaluation of velocity and displacement with no apparent numerical instability.

The relative velocity in Figure 4 shows some unusual behaviour. However, this is explained by the behaviour of the damping equation (Equation 3). In particular, as the relative displacement begins to exceed 1 mm, the damping increases drastically, causing a sudden and strong restoring force. This explains the relative speed spikes at approximately 0.08 and 0.13 s. Repeating the simulation over different cases further illustrated the presence of a non-linear response in the skin and the implied existence of an inflection point below which it may be reasonable to model the skin as linear. Since the relative displacement shown in Figure (weaker hit) does not exceed 1 mm, no such spike in relative velocity was observed and the non-linear model effectively matched the linear model.

The parametric sweep study showed that there is a clear point of inflection at which the non-linear model begins to diverge significantly from the linear model. For the *t* = 0.01 case (Figure 6), this point occurred at an input magnitude of 272 $m \cdot s^{-2}$ (Figure 6). Below this magnitude, the non-linear model's peak acceleration output followed a somewhat linear path, similar to the linear model. Figure 6 also shows that for smaller duration impacts, the inflection occurs at larger acceleration magnitudes. However, when inflection occurs, divergence from the linear model was more pronounced. It is therefore reasonable to accept that, for accelerations below the inflection points, it may be valid to use a linear model when simulating this STA. However, even in these cases, a degree of bias will affect the results due to the non-linear nature of the physical system. This is illustrated by the already diverging values at accelerations lower than the point of inflection.

This model for the skin does not account for tensile or compressive failure of the skin, though it is unlikely that this would have a major effect on the outcome at such small displacements. It would also be exceedingly difficult to collect such data from *in-vivo* skin in an ethical way. In contrast, the mechanical behaviour of skin *ex vivo* has commanded considerable study (Joodaki & Panzer, 2018), including the high strain rate behaviour (Butler et al., 2015; Shergold, Fleck, & Radford, 2006) and the local composition and directionality effects (Khatyr, Imberdis, Vescovo, Varchon, & Lagarde, 2004). Although there are known degradative effects on mechanical properties of *ex vivo* biomaterials (with experimental methods to minimise such effects continually improving), strain-rate effects are still widely acknowledged in skin. Therefore, viscoelastic properties of skin need to be considered during a dynamic event such as those outlined here.

Parker et al (2016) presented data showing the nonlinear nature of stiffness and damping over a transverse displacement range similar to the deviations of the skin mounted sensors. Thus, the experiment enabled development of an empirically justified non-linear model of coupling between the skull and the skin/sensor. Due to ethical and instrumentation limitations, *in vivo* measurement of the discrepancy between peak accelerations taken from sensors coupled with the skull and skin mounted sensors is difficult. However, the approach in this study is validated axiomatically by providing a model that captures the disparate observations in peak acceleration measured by skin mounted sensors King et al., 2017; King et al., 2016) and skull mounted sensors (Bartsch et al., 2019). Furthermore, it shows that the relative equivalence in skin mounted sensors and skull mounted sensors at low impacts (Wu et al., 2016) can be explained by the non-linear skin mechanics.

The simulation did not account for relative motion between the skull and the accelerometer components within the mouthguard. However, the coupling between the mouthguard and the player's teeth was tight and unlikely to significantly affect the results, when compared to the dominating non-linearity of the skin. The use of skin-mounted inertial sensors is common in sport-related head impact research (Chrisman et al., 2016; Hecimovich, King, Dempsey, Gittins, & Murphy, 2018; King, Hecimovich, Clark, & Gissane, 2017; Doug King et al., 2017a; D. King, P. Hume, C. Gissane, & T. Clark, 2017b; D. A. King, P. Hume, C. Gissane, & T. Clark, 2017; Lynall et al., 2016). Previous in-vivo experiments have shown that skin-mounted inertial sensors can over-estimate head accelerations by 120% (Wu et al., 2016). The model presented in Equation 1 predicts that a skin-sensor system yields a much greater over-estimation in the peak acceleration of the head during a harsh sporting head impact. The discrepancy in over-prediction between this model and the experiments of Wu et al. can be explained by the relatively low magnitude impacts (~15 g) used in their *in vivo* study, due to the ethical concerns surrounding intentional impacts. Figure 6 shows that 15 g (147 $m \cdot s^{-1}$) impact is below the inflection points of the non-linear model. The present simulations demonstrate that the errors do not diminish with larger magnitude impacts. Rather it implies the discrepancies due to STA may increase with impact magnitude.

This discrepancy has important implications for the interpretation of existing sport-related head impact data. In general, the median (or mean) head impact accelerations reported across a variety of sports are consistent and relatively low, ranging from 13 – 22 g research (Chrisman et al., 2016; Hecimovich et al., 2018; D King et al., 2017; Doug King et al., 2017a, 2017b; D. A. King et al., 2017; Lynall et al., 2016). That is only marginally higher than the 10 g threshold usually employed to prevent sensors from recording motion associated with daily activities like running (King, Hume, Gissane, Brughelli, & Clark, 2016; Zhang et al., 2004). However, the maximum impact magnitudes reported in these studies are high, ranging from 66 – 153 g. Based on the simulation results, it is likely that these values are over-estimated by approximately 160%. Similarly, existing brain injury thresholds may substantially overestimate the head accelerations associated with sport-related concussion (Campolettano et al., 2020; Mihalik, Lynall, Wasserman, Guskiewicz, & Marshall, 2017; Zhang et al., 2004). These thresholds have been determined based on head accelerations recorded using instrumented helmets, which are subject to even greater errors than skin-mounted patches (Wu et al., 2016). Therefore, they should be used with caution when monitoring the limits of impact tolerance regarding player health. Head accelerations recorded with instrumented patches and helmets are unlikely to provide realistic input to finite element simulations of brain injury (Beckwith et al., 2018). Significant discrepancies can occur when using differing sensor systems that have variable coupling tolerances with the skull. To get the best estimates of skull motion, data should be collected using the most tightly coupled inertial sensors.

CONCLUSIONS

In this study, empirical skin mechanics data from Parker et al (2016) was used to develop a model of the coupled skull-skin-sensor unit system. The model was used to investigate impact of STA on the overestimation of head acceleration inferred by skin mounted sensors during large impacts. It was observed that when modelled as a physiological, non-linear system, a sensor mounted on the skin can substantially overestimate skull accelerations. These findings are in agreement with results published in existing literature and may explain the discrepancy between skin- and mouthguard-mounted accelerometer measurements. Future head impact telemetry studies should ensure secure sensor-skull coupling of skull motion measurement devices to minimise the effects of soft tissue artefact, ensuring data integrity.

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