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# On the ability of experimental impact measures to predict tooth injuries in an *ex vivo* swine model

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## Abstract

**Background/Aim:** Impact to the orofacial region, in particular teeth, is a frequent incident leading to injury in many sports and can result in health and economic costs for the injured individual. The majority of previous work has applied synthetic models such as plaster or stone, to form analogs of relevant structures to study the potential for impact-induced injury. Biomechanical studies that have applied tissue models (animal or human) for the purpose of determining the biomechanical measures associated with dental injury are rare. The aim of this study was to apply a simple *ex vivo* model based on swine dentition to ascertain which of a select list of measurable quantities associated with impact mechanics could predict luxation and fracture of teeth due to impact.

**Methods:** Mandibular central incisors of *ex vivo* swine dentitions were impacted using a linear drop tower with heights ranging from 1.20 m to 2.42 m. Seven mechanical predictors were assessed at impact and were then subjected to binary logistic regression techniques to determine which was the best predictor of luxations or fractures of the teeth.

**Results:** Of the seven mechanical predictors, (1) the velocity of the impacting body ($R^2$ = 0.477), (2) a proxy measure for the change in kinetic energy of the impacting body ($R^2$ = 0.586), and (3) the approximate energy absorbed by the tissue ($R^2$ = 0.722) were found to be statistically significantly different ($p < .05$), offering the greatest specificity as indicated by receiver operator characteristics. Other measures that are frequently used in impact mechanics, including peak linear acceleration and velocity change, were not statistically significant predictors of tooth injury.

**Conclusion:** Identifying mechanical predictors for dental injury of unprotected teeth provides a first step in understanding which aspects of an impact event attribute to dental injury and can lay the foundation for future studies that examine alteration in injury mechanics associated with protection devices.

KEYWORDS
dental injury, mouth protection, tooth fracture, tooth luxation, trauma biomechanics

## 1 | INTRODUCTION

Impact to the orofacial region and, in particular impact involving the teeth of both the maxilla and mandible, can lead to dental injuries including fractures, several types of luxations, and avulsions.[1–5] Common causes of these injuries are associated with sports, violence, motor vehicle accidents, and falls.[1,6] In youth, impact associated with involvement in sports is a frequent cause of dental injury.[1,7] In the United States between 1990 and 2003, there was an annual average of 22,000 sport-related dental injuries in people aged less than 18 years.[8] In a study spanning a decade and considering 3385 craniomaxillofacial trauma cases, 31.8% of injuries were associated with sport.[9] Dental injuries related to sport are associated with lifetime economic costs estimated at $20,000 for a single injured individual.[10]

Mouth protectors, sometimes referred to as mouthguards, are the primary protection devices used by both youth and adults in sports with the risk of impact-induced dental injury. Despite the various classifications of mouth protectors, all mouth protectors encase the teeth of the maxillary arch and certain portions of the gingiva.[11,12] Viscoelastic polymers are a typical protective material that provide protection against impact-induced injuries through two phenomena: (1) energy dissipation and (2) splinting.[1] Both energy dissipation and splinting are thought to combine to reduce impact forces on teeth, relative to forces that would be experienced if the teeth were not encased. While mouth protectors are thought to be capable of reducing the likelihood of dental injury, it is important to acknowledge that there is limited biomechanical research that specifically examines the mechanics of tooth injury using *in vivo* or *ex vivo* dentitions.

Experimental work on impact-induced dental injury mechanics is focused on the application of synthetic models. The models include relevant oral structures and report various mechanical metrics as an evidence base for the protection ability of mouthguards or proxy measures for injury. Godwin and Craig[13] described a synthetic maxillary model coated in brittle lacquer and examined the appearance of cracks in the lacquer after impact to ascertain protection. In later work, these authors again used stone models of dentitions to examine the ability of several mouth protector types and constructions to attenuate impact energy delivered by an impact pendulum.[14] Oikarinen and colleagues[15] used plaster models of the maxillary dentition, subjected to impact using a falling impactor resembling an ice hockey puck, and recorded a proxy for impact energy needed to cause fracture in the plaster. Greasley and colleagues[16–18] applied a model dentition comprising a composite jaw, rubber arch, and dental stone models of teeth claimed to possess fracture toughness matching select *ex vivo* and *in vivo* data. The fracture of the jaw or teeth (due to impact) as indicators of injury was reported. Hoffman and colleagues[19] applied a model where teeth deflected from labial to lingual directions, due to impact forces, and quantified the percentage reduction in the deflection as an indicator of protection efficacy. Synthetic models lack tissue characteristics (either *ex vivo* or *in vivo*), where soft tissue such as the periodontal ligament (PDL) between the tooth and alveolar bone are present. Tissue characteristics are necessary in order to properly assess for PDL damage and ultimately establish which mechanical parameter best predicts dental injury. Verissimo and colleagues[20] implemented a non-destructive *ex vivo* bovine model in which the incisors were subjected to impacts from a pendulum. A strain gauge located on the opposite side of the tooth from the impactor measured strain during impacts. Using a finite element model, it was concluded that the presence of a mouthguard decreased the stress and strain in the incisor, which is likely to reduce injury. Patterson and Popowics[21] applied intrusive traumatic impacts to *ex vivo* swine central incisors and reported the displacement of the root concluding the model was appropriate to study traumatic damage to the periodontal support but it did not consider the mechanics of the applied impact. Injury mechanics specific to the tooth were investigated by Pilo and colleagues[22] where a bursting pressure was applied to extracted human incisors and premolars to determine the tooth strength. Although providing useful mechanics for the tooth, this method does not consider the PDL support, alveolar bone or the mechanics describing the impact event that precipitates the potential dental injury.

Because these studies provide limited information on the mechanics specifically related to the dental injury event, it is reasonable to question whether the reported measures, such as the reduction of impact energy or tooth deflection, are justifiably proportional to a reduction in injury. Without an understanding of the mechanical parameters associated with dental injury, there is no way of determining which mechanical parameter a mouth protector should be designed to attenuate to be effective. Establishing an appropriate set of mechanical parameters that can predict dental injury for unprotected teeth is an essential first step toward understanding impact-induced injury. To the authors' knowledge, there are no biomechanical studies that have applied tissue models (animal or human) for the purpose of determining appropriate biomechanical measures to describe a dental injury event. Therefore, the aim of this study was to use an unprotected swine *ex vivo* model and establish, using statistical approaches, the mechanical measures most predictive of dental injury within a Tooth-PDL-Bone-Complex (TPBC). A binary logistic regression technique was used to examine whether impact acceleration, impact speeds, and estimates of the energy absorbed by the impacted dentition were possible predictors of dental injury. Determining such predictors can (1) provide knowledge on the appropriate mechanical parameters directly attributable to dental injury and (2) lay the initial foundation for an evidence-based standard or criterion on which to assess the impact protection mechanics for mouth protectors.

## 2 | MATERIALS AND METHODS

Swine mandibles ($n$ = 22) were obtained fresh-frozen from a slaughterhouse. The swine used for this study were sacrificed for food purposes prior to the present study, and therefore, ethics exemptions were granted. Mandibles were used, as opposed to a maxilla,

because the maxillary dentition was deemed excessively worn and damaged, while the mandibular dentition, in particular the central incisors, was intact and of comparable size across the obtained specimens. The focus of the simplified swine mandible model presented is the mechanics of the TPBC with respect to dental injury. Other risk factors and interactions such as those from occlusion and overjet were not considered. In an effort to assess the range of physical size of the incisors and approximate PDL thickness, mandibles that did not experience significant damage due to fracture were imaged post hoc using Micro-Computed Tomography (microCT) scans (U-CT, MILabs BV, current 0.19 mA, voltage 55 kV and reconstructed with a voxel size of 40 micron) and morphological measurements were made using software (Avizo v9.1.1, Thermo Fisher Scientific) (Figure 1). The resulting average, standard deviation, and coefficient of variation (ratio of the standard deviation to the average) of each morphological measurement were calculated. In addition, the mass of each mandible and dental stone base was recorded and the average, standard deviation, and coefficient of variation were calculated for the mandibles with and without injury.

A custom-built monorail drop tower was used to impact the swine model (Figure 2). An impactor that was guided through free-fall by a linear bearing consisted of a ball arm with a rounded 3-inch thick piece of vulcanized rubber (standard ice hockey puck) secured to the impacting end. The acceleration of the impactor was measured with a uniaxial accelerometer within the ball arm (Neill-Tech model CAV2147V, Xiamen Neil Electronics Co., 2000 *g* range, 0.2 mV/g sensitivity) during free-fall and impact. The distal most portion of the mandible was cast in dental stone and secured in a spring-anvil. The spring-anvil allowed downward vertical travel during the impact event as opposed to a rigid boundary condition. This allowed the dentition to translate in a manner roughly approximating the *in vivo* gross head motion. However, it did not account for control of the dentition from the temporomandibular joint and associated musculature that are present *in vivo*. A tri-axial accelerometer measured acceleration of the mandible-stone assembly (Adafruit Industries ADXL377 LLC part # 1528-1018-ND, 200 g range, 6.5 mV/g sensitivity). Each impact was recorded using a high-speed camera, collecting images at

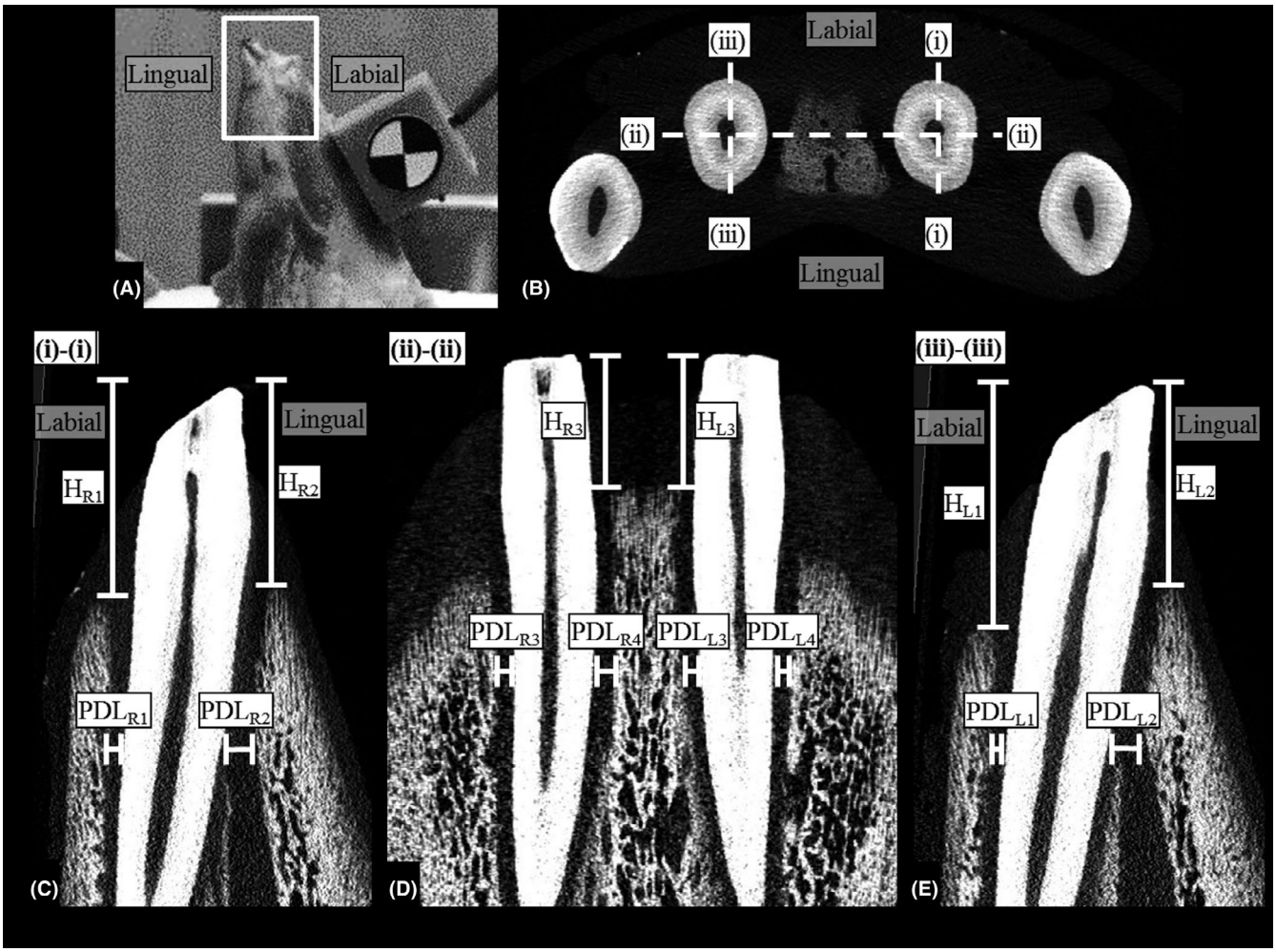


**FIGURE 1** (A) A representative view of the mandible with both the relevant anatomical directions indicated and the area of the mandible in which the MicroCT images were taken. (B) A top view of the incisors indicating the planes at which anatomical measurements were taken (C) Right incisor in plane (i)-(i) with indication of the measurements taken. (D) Both left and right incisors in plane (ii)-(ii) looking toward the lingual side, with indication of the measurements taken (E) left incisor in plane (iii)-(iii) with indication of the measurements taken

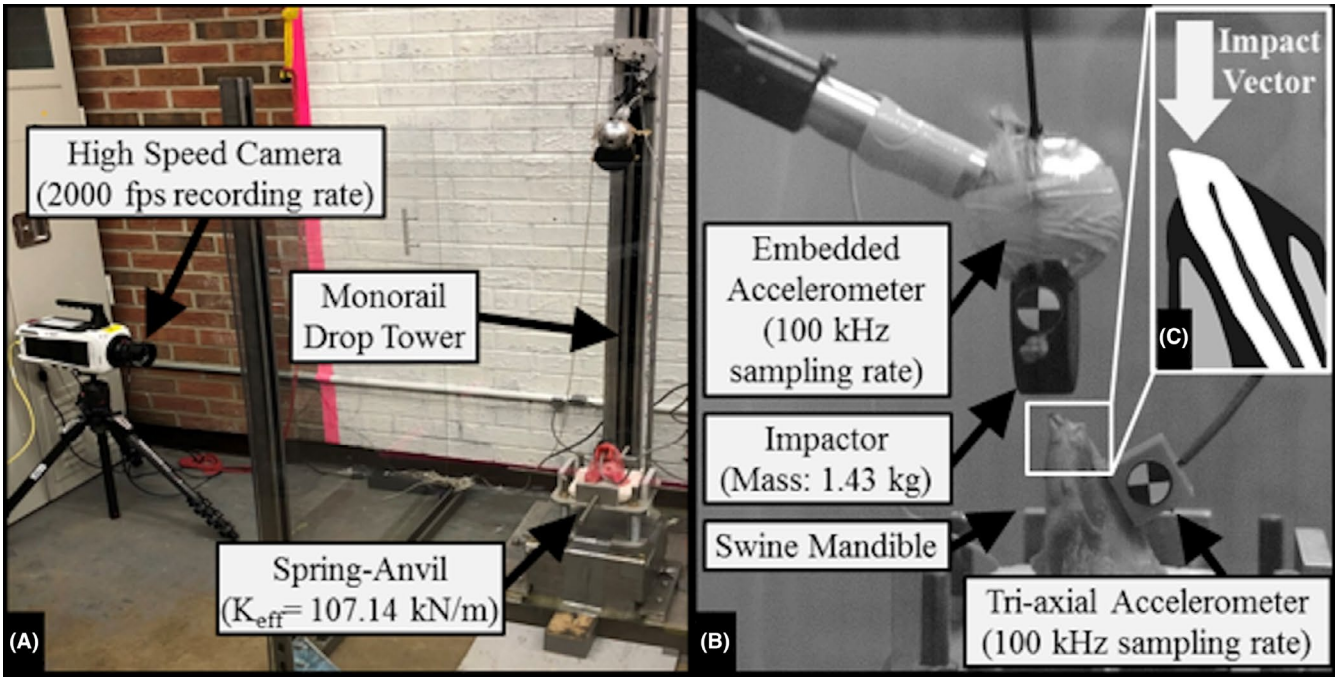


FIGURE 2 (A) Experimental setup including the custom-built monorail drop tower with a swine mandible secured within the spring-board anvil at the base and the high-speed camera setup to capture the impact. (B) Image of the swine mandible and impactor and the location of accelerometers. (C) Representative cross section view of the central incisors with the approximate impact vector

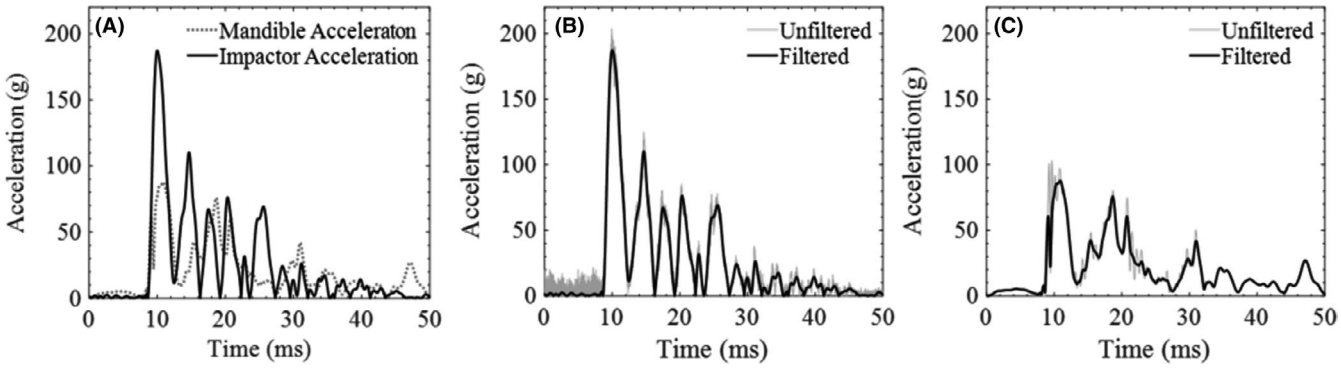


FIGURE 3 (A) Example data for the accelerations recorded from both the accelerometer instrumented within the impactor and fixed onto the mandible during impact. (B) Example data showing the unfiltered and filtered data for the accelerometer instrumented within the impactor, (C) example data of the unfiltered and filtered data for the accelerometer fixed onto the mandible

2000 frames per second and equipped with a 50 mm f/1.4 lens (Phantom v611, Vision Research Inc.). The acceleration data from the impactor and mandible were collected and saved at 100 kHz using National Instruments hardware and software (PXI 6251 and Labview v18.0.1). Analog voltages from the uniaxial impactor accelerometer were anti-alias filtered by a hardware low-pass filter with a cutoff frequency of 4 kHz. Measured data from both accelerometers were also filtered in post-processing with a low-pass filter at 1 kHz (CFC 600), (Figure 3).[23]

Each mandible was subjected to one impact. The height of the impactor above the exposed incisors was varied, ranging from 1.20 m to 2.42 m, and injury was assessed after each impact. The occurrence of tooth fracture was evident visually and tooth luxation was confirmed by visual observation of displaced teeth or manual palpation after impact.

A total of seven mechanical parameters were determined as variables describing the impact event: velocity of impactor before impact (Figure 4, $V_i$, calculated from high-speed video, pre-impact velocity is positive valued, m/s), average velocity of the mandible post-impact (m/s), peak acceleration of the impactor (g, where 1 g = 9.8 m/s/s), peak resultant acceleration of the mandible (g), change in velocity of the impactor (rebound velocity, $V_f$, of the impactor is negative valued, m/s), change in the square of the impactor velocity (pre-impact to post-impact, $m^2/s^2$), and a calculated approximation of energy absorbed by the mandible (J) that required determination of the maximum displacement of the

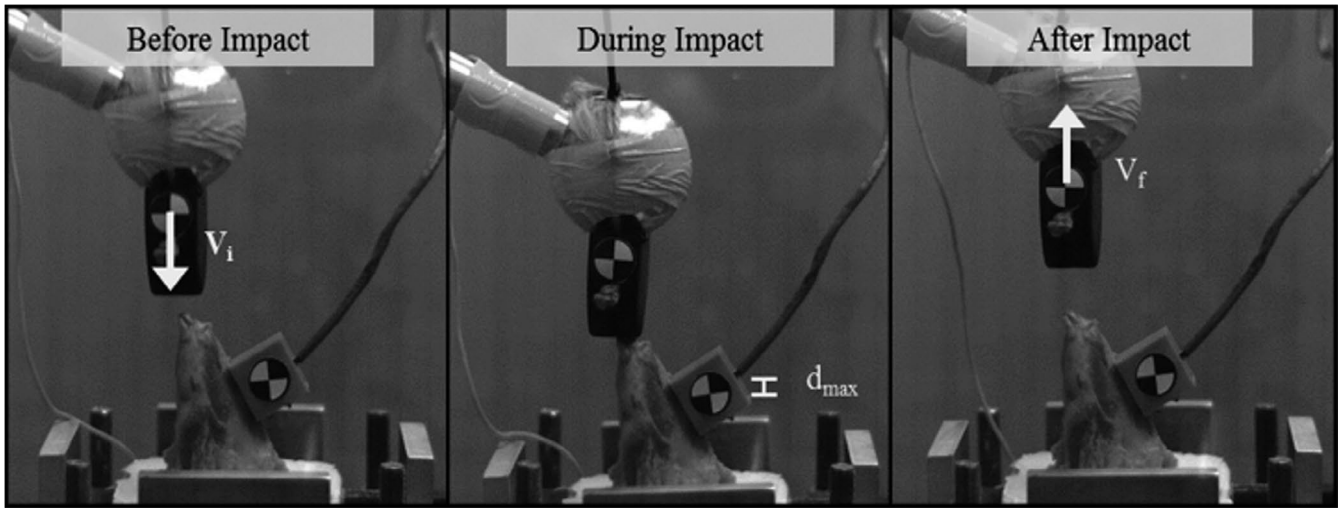


**FIGURE 4** Images from high-speed video representing three time stamps from before, during, and after impact. From left to right, the variables indicated illustrate the velocity of the impactor before impact, the maximum displacement of the mandible during impact and the velocity of the impactor after impact

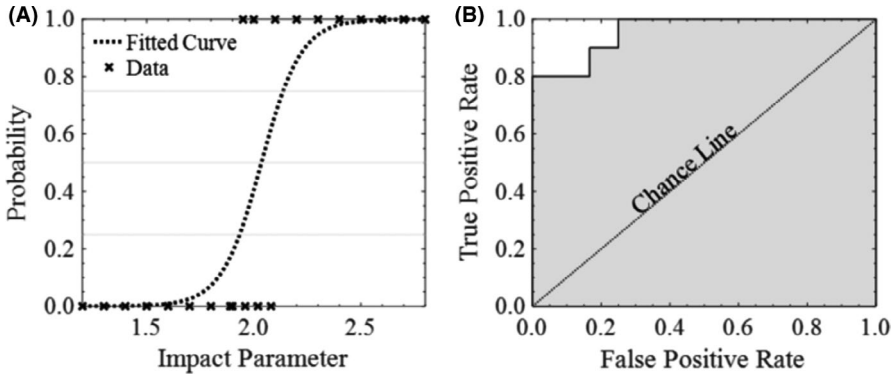


**FIGURE 5** (A) Exemplar binary output results and a regression fit for a logistic function that can be interpreted as probability of injury for the impact parameter magnitude (not specified in the example plot shown) in the specific injury model. Grid lines correspond to 25%, 50%, and 75% probability. (B) Exemplar ROC curve, where the line demarking the un-shaded and shaded regions is referred to as the ROC curve. The area that is shaded is referred to as the area under the ROC and in general, areas that range from 0.5 to 1.0 indicate the parameter is as predictive as chance (0.5) toward ability to perfectly distinguish between outcome state (1.0), respectively

mandible during impact $d_{max}$ (ie, the displacement of the 4 springs, Figure 4), as shown in Equation (1):

$$\text{Predicted Energy Absorbed} = \frac{1}{2}\left[m_{\text{impactor}}\left(v_i^2 - v_f^2\right) - k_{\text{eff}} d_{max}^2\right] \quad (1)$$

For each mechanical parameter, a binary logistic regression was used to predict experiment outcome (eg, injury—code 1 or no-injury—code 0). This amounted to seven binary logistic regressions that were performed where each mechanical parameter was considered a predictor. Prior to performing each binary logistic regression, the assumption of linearity and outliers was tested, using established techniques.[24,25] A Bonferroni adjustment was applied in the test by assigning a three-parameter model (a mechanical parameter, an interaction term, and a constant). Linearity was assumed when $p > .0167$ ($\alpha = .05$ divided by 3 terms). Outliers were defined as data showing standardized residual greater than ±2.5 standard deviations. For each regression model, the Nagelkerke $R^2$, sensitivity, specificity, receiver operator characteristic (ROC), the statistical significance of the mechanical predictor in the regression model (significance level of $\alpha = .05$), and the probability of injury curve were reported. An example showing key statistical analysis outputs for the regression is shown in Figure 5. To determine an appropriate sample size for this study, an event per variable guideline (EPV) of 10 proposed by Peduzzi and colleagues[26] was used. The guideline states that if the lowest experimental outcome or event (injury or no-injury) amounts to at least 10 cases, one predictor variable may be used per each logistic regression model. To conform with this guideline, the study ensured at least 10 events of injury and no-injury occurred out of the 22 swine mandibles.

## 3 | RESULTS

From the 22 impacted swine mandibles, 10 had tooth luxations or fractures while the other 12 did not have these injuries. Key anatomical dimensions are noted in Table 1, the PDL dimensions $R_2$ and $L_2$ had the greatest coefficient of variation across all specimens, 74% and 68%, respectively, while the majority of the other metrics in Table 1 had approximately 20% coefficient of variation and less. The average mass of each mandible and dental stone base for the injured and non-injured trials is shown in Table 1 (no-injury cases 0.80 ± 0.14 kg and injury cases 0.79 ± 0.12 kg), and these indicate that the mass and inertial properties of the injured and non-injured models are comparable. An exemplar high-speed video of an impact inducing a fracture injury can be found in the supplementary materials (Supplementary Video S1).

The binary logistic regression analysis (summary data presented in Table 2) reports impactor velocity (an experiment input condition, m/s), change in the square of the impactor velocity ($m^2/s^2$), and approximated energy absorbed by the mandible (J) as statistically significant predictors of tooth injury ($p < .05$). Approximated energy absorbed by the mandible had the greatest Nagelkerke $R^2$ (0.722) and ROC area (0.925). All of the mechanical predictors were found to be linearly related to the logit of the tooth injury dependent variable ($p > .0167$). Standardized residuals with a value of 3.08, 3.55, and 2.51 standard deviations were found for velocity before impact, change in the square of the impactor velocity, and approximated energy absorbed by the mandible, respectively. These standardized residuals were therefore considered outliers (greater than ±2.5 standard deviations) but the measured data leading to these residuals remained in the analysis.

Binary logistic curves conveying probability of injury, and ROC curves for each of the three significant mechanical predictors are presented in Figure 6. As shown, the presented mechanical predictors were more predictive than chance and the approximated energy absorbed had the greatest area under the ROC.

**TABLE 1** Mean, standard deviation, and coefficient of variation for the mass of each mandible including the dental stone base ($n$ = 22), and the anatomical measurements for the central incisors of the mandibles ($n$ = 14)

| | Mean ± SD | Coefficient of Variation |
|---|---|---|
| Mass No-Injury (kg) | 0.80 ± 0.14 | 0.17 |
| Mass Injury (kg) | 0.79 ± 0.12 | 0.15 |
| PDL*xx* (mm) | | |
| R1 | 0.73 ± 0.12 | 0.16 |
| R2 | 1.17 ± 0.86 | 0.74 |
| R3 | 0.78 ± 0.12 | 0.15 |
| R4 | 0.80 ± 0.16 | 0.20 |
| H*xx* (mm) | | |
| R1 | 12.26 ± 1.12 | 0.09 |
| R2 | 11.92 ± 1.45 | 0.12 |
| R3 | 7.25 ± 1.66 | 0.23 |
| PDL*xx* (mm) | | |
| L1 | 0.76 ± 0.18 | 0.24 |
| L2 | 1.01 ± 0.69 | 0.68 |
| L3 | 0.78 ± 0.16 | 0.21 |
| L4 | 0.78 ± 0.13 | 0.17 |
| H*xx* (mm) | | |
| L1 | 11.97 ± 1.41 | 0.12 |
| L2 | 11.83 ± 1.36 | 0.11 |
| L3 | 7.00 ± 1.46 | 0.21 |

Anatomical measurements were not included for eight mandibles due to significant fracture damage making post-impact imaging not feasible.

## 4 | DISCUSSION

In the present work, both luxation and tooth fractures were created in the central incisors of the swine model. Three-dimensional imaging techniques through microCT reconstructions were used to ensure that the mandibles used in the study contained dimensionally similar incisors and PDLs. Given that the expected potential of the teeth to absorb impact energy will be based in part on dimensions of these structures, assessing dimensional similarity across the specimens was a prudent check that assured uniformity in the swine model. The output measures from the impact experiment best predicting injury were (1) approximated energy absorbed by the mandible followed by (2) the difference of the squares in pre-impact and post-impact velocity of the impactor (a proxy measure for change in kinetic energy of the impactor pre-impact to post-impact). This work adds to the scant literature examining impact-induced injury to the dentition using an *ex vivo* swine model. Indeed, the majority of previous reports apply synthetic *in vitro* models, typically plasters, stones, and lacquers modeling the shape of the human dentition as opposed to biological tissue that can exhibit luxation and fractures. Furthermore, the majority of these previous studies have focused on examining the mechanics of dental protection offered by polymer mouth protectors without first establishing the mechanics that predict impact injury. Arguably, establishing the mechanics most predictive of injury in unprotected teeth is a prudent first step that could form a basis upon which later investigations of protection mechanics could build.

Biomechanical investigations of impact-induced injury that are structured toward ascertaining which mechanics are predictive of injury must consider a wide array of predictor variables, each with plausible links to the injury event considered. The present work considered six output measures and one input measure (impact velocity $V_i$) from the impact experiment. Impact velocity was investigated as a plausible predictor as it is controllable and it has previously been reported that with a spring-mounted simulated maxillary arch, increased impact speed led to an increase in the number of teeth broken during impact.[18] Various measures of velocity change during impact can allow estimation of, and have

**TABLE 2** Summary report for statistical analyses of impact measures when regressed against injury outcome

| Mechanical Predictor | Nagelkerke $R^2$ | Sensitivity and Specificity | ROC Area | ROC 95% confidence interval | $p$-value with $\alpha = .05$ |
|---|---|---|---|---|---|
| Impactor velocity (m/s) | 0.477 | 90.0% and 75.0% | 0.875 | 0.726 to 1.000 | .033 |
| Change in the square of the impactor velocity ($m^2/s^2$) | 0.586 | 90.0% and 83.3% | 0.908 | 0.779 to 1.000 | .030 |
| Predicted energy absorbed by mandible (J). | 0.722 | 80.0% and 83.3% | 0.925 | 0.812 to 1.000 | .041 |
| Change in velocity of the impactor (m/s) | 0.008 | 20.0% and 83.3% | 0.525 | 0.273 to 0.777 | .709 |
| Average velocity of mandible (m/s) | 0.064 | 50.0% and 66.7% | 0.608 | 0.366 to 0.851 | .309 |
| Peak acceleration of impactor (g) | <0.001[a] | 0.0% and 100% | 0.483 | 0.232 to 0.735 | .950 |
| Peak acceleration of mandible (g) | 0.047 | 40.0% and 66.7% | 0.575 | 0.328 to 0.822 | .383 |

[a]The peak acceleration of the impactor Nagelkerke $R^2$ was found to be 0.000239.

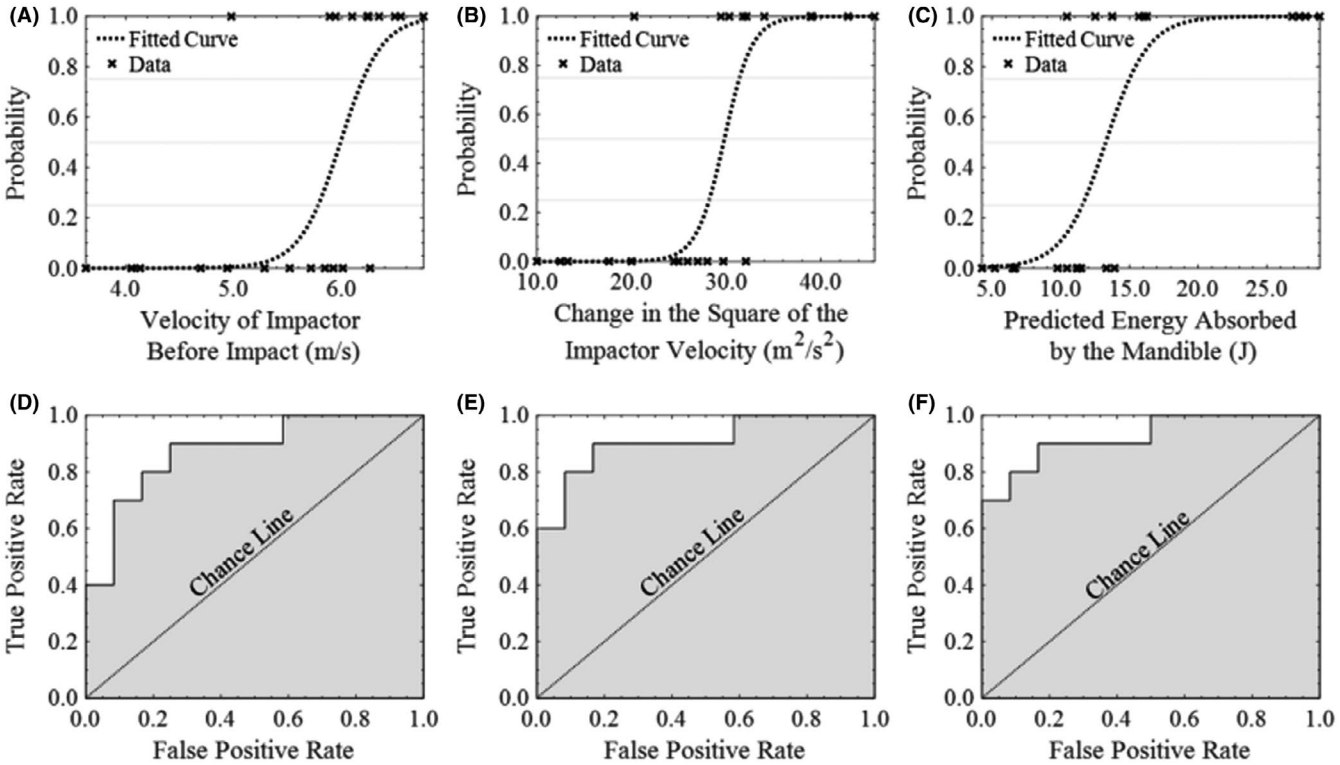


**FIGURE 6** Probability of injury curves and ROC curves for the three statistically significant predictors. (A–C) The probability of injury curves with respect to the velocity of the impactor before impact, change in kinetic energy of the impactor and predicted energy absorbed by the mandible, respectively. Grid lines correspond to 25%, 50%, and 75% probability. (D–F) The ROC curves for the velocity of the impactor before impact, change in kinetic energy of the impactor, and predicted energy absorbed by the mandible, respectively

been invoked previously either directly or indirectly in, functionals for injury likelihood estimation in automotive crashworthiness testing.[27] The change in velocity of the impactor and the velocity of the dentition after impact can be interpreted as proxy measures for the change in momentum during impact, which is dictated by the stiffness of the dentition and the spring-anvil system.[18] Traumatic loading to the periodontal support has been shown to decrease the stiffness of the TPBC system,[21] arguably altering the change in momentum. The ANSI/ADA Standard No. 99 requires 65% energy absorption in a protected mandible,[11] and thus investigation into attainable proxy measures for energy absorption as predictive of dental injury is crucial. Approximation of the energy absorbed by the dentition and the change in the square of the impactor velocity, a proxy measure for the change in kinetic energy of the impactor, were utilized for this. Peak accelerations are typically interpreted as proxy measures of impact force and are often used to quantify impact attenuation in the assessment of protection equipment including protective headgear.[28–30] In past biomechanical literature focusing on skull fracture and severe life threatening brain injury, peak accelerations have appeared as measures on which injury likelihood can be inferred.[31,32] Based on the findings of previous work surrounding impact-induced injury, the seven predictors for injury considered in the present work are

arguably logical choices for a first assessment of impact-induced dental injury.

Experimental output measures, acceleration (units of g), and velocity change ($V_i$–$V_f$, where velocity pre-impact is a positive quantity and post-impact is a negative quantity, m/s) were the poorest predictors of injury. It is speculated that this is a result of the mechanics associated with tooth luxation and fracture. One might expect that as drop height (and impact velocity) increases, accelerations, and velocity change of the impactor and mandible would scale to increasing magnitude because the forces experienced by both impactor and dentition increase in direct proportion to drop height. This could be the case for a model that did not exhibit tooth fracture and that is mechanically rigid. For example, head models used in head injury assessment that are not designed to fracture typically measure increasing acceleration and velocity change as impact velocity increase.[33] In contrast, in a model that fractures (or, as in this study, also exhibit luxation), fractures and luxation can manifest as mechanical compliance. In such a model, as drop height (impact velocity) increases, one might expect increases in acceleration and velocity change up to the point where the drop height (impact velocity) is great enough to create fracture or luxation. Once fracture and/or luxation occur, accelerations and velocity change might no longer increase but instead be dictated by complicated fracture mechanics.

The experimental outputs associated with energy change of the falling impactor, and the approximating Equation (1) for energy absorbed by the swine mandible, were statistically significant predictors that exhibited the greatest Nagelkerke $R^2$ and area under ROC. In future work, protected teeth will be subjected to impact and using the same regression techniques, the experimental outputs will be examined to determine whether the mechanical measures can predict injury likelihood and therefore detect a protective effect of mouth protectors. The research questions in such a follow-on study, building on this present work, will seek to answer whether (1) change in the square of impactor velocity and approximate energy absorbed are statistically significant predictors of injury to protected teeth; and (2) whether or not the magnitudes of these parameters increase or decrease in protected teeth relative to the magnitudes found in the present work focusing on unprotected teeth.

While the authors are not aware of previous reports using an *ex vivo* model matching the one described in this study, a previous report applying spring-mounted synthetic models of the dentition could allow for qualitative assessment of how the present results compares to previous studies.[16] Greasley and Karet[16] applied a synthetic model of the jaw and teeth, including a soft junction between jaw and teeth, and claimed that rounded conical impactors (5 mm tip radius) falling at 6.25 m/s, corresponding to 10 Joule energy, created clinically relevant injury patterns (as indicated by damage to the model). They found that such a model was able to detect the presence of mouth protection (as evidenced by reduced number of broken teeth with guarded teeth). In the present work, injuries were noted at 5–6 m/s velocities (and above), corresponding to 17.9 Joule (at 5 m/s), using a rounded rubber impactor with a radius of approximately 37 mm. The spring support applied by Greasley and Karet[16] is reported to have 1 kg/mm stiffness, while the present work applied 10 kg/mm stiffness. Despite differences in spring support and model, it could be argued as encouraging that the *ex vivo* swine model exhibits injuries at speeds and energies comparable to selected previous work.

Like all *ex vivo* studies, the present work has limitations. The swine model was chosen to allow for *ex vivo* investigation using tissue that did not exhibit the variance and degradation typically associated with post-mortem tissue from donors of advanced age. Nevertheless, it is acknowledged that there are mechanical differences between the swine dentition and humans. It is speculated that the present swine model is an appropriate one to examine which mechanics could predict injury in the TPBC and it could be applied in future work. The present study focused on the TPBC interaction leading to dental injury and is simplified from an *in vivo* injury event. The simplified boundary conditions induced by the spring-anvil and isolated mandible do not include all relevant anatomical variations that can affect injury risk factors such as musculature, overjet, occlusion, and interaction from the temporomandibular joint. The simplification used here provided a repeatable means for securing the mandibular specimens and control of experimental variables (eg, alignment) to study local dental trauma on swine mandibular incisors. As such, baseline data have been provided on which future work can expand to consider how additional factors (eg, degrees of freedom and compliance of the temporomandibular joint) affect injury events. As the morphology and alignment of mandibular teeth may vary from that of maxillary teeth, and the mandible is joined to the skull through musculature and the temporomandibular joint, further work may be necessary to determine if, and how, the magnitude of outcome measures differs between the two arches. Future work could consider a more comprehensive model that contains complex interactions and more sophisticated injury identification approaches. The impact experiment, involving a hockey puck and incisors, is a model of only one of the many possible scenarios that could result in dental injury.

In part due to a lack of availability of historic data, it was not possible to do an a priori sample size determination via power analysis. To date, there has been no previous biomechanical work that has used binary logistic regression techniques to predict dental injury. Therefore, previous knowledge of a mean, standard deviation, odds ratio, or the probability of dental injury for the mechanical predictors was unknown, all of which would have been required to perform a reasonable sample size calculation. Nevertheless, an EPV of 10 was applied to guide the sample size of the current study as recommended from previous literature.[26] It is acknowledged that the findings could be limited to the simplified impact scenario examined, but the puck-incisor impact is a plausible scenario relevant to sport carrying significant risk of injury.

The findings from this preliminary study determined the approximated energy absorbed by the mandible followed by the difference of the squares in pre-impact and post-impact velocity of the impactor (a proxy measure for change in kinetic energy of the impactor

pre-impact to post-impact) were the most effective predictors of injury outcome.

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## AUTHOR CONTRIBUTIONS

All listed authors participated in study design and critical review of manuscript content. Alexiou and Graf provided equipment access and critical analysis of imagery data. Adanty, Houg, MacGillivray, McAllister, and Dennison lead data collection and analysis.

## CONFLICT OF INTEREST

None of the authors have a personal or professional relationship with any of the equipment vendors for research tools listed and do not stand to gain financially by the publication of the research work.

## DATA AVAILABILITY STATEMENT

The data that support the findings of this study are available on request from the corresponding author.

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## SUPPORTING INFORMATION

Additional supporting information may be found online in the Supporting Information section.

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