Finite Element Analysis of Maxillary Central Incisor Trauma

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ABSTRACT

JORDAN L. OLSEN: Finite Element Analysis of Maxillary Central Incisor Trauma
(Under the direction of Drs. Lorne Koroluk)

Biomechanics of traumatic dental injuries (TDI) is not well understood. Computer-based simulation/finite element method—FEM may facilitate understanding TDI. A realistic 3-D FEM of a maxilla with multi-tooth loading will enhance TDI understanding.

METHODS: FE model of a complete maxilla including alveolar process, enamel, dentin, pulp, and PDL was constructed using CT imaging of an 18 year old female. Single and multi-incisor TDI was simulated using horizontal, oblique, and vertical loading with static point forces of 200 N (right central incisor) and 400 N (both central incisors). Stress/strain/displacement of affected/adjacent teeth and supporting structures were evaluated based on TDI scenario. RESULTS: Horizontal loading experienced most stress in all hard/soft tissues compared to other loading scenarios. Double loading experienced approximately double stress/displacement. Teeth in isolation to other structures experienced most stress while PDL experienced little. CONCLUSION: Anatomically accurate full maxilla FE model was constructed. FE model output approximates clinical presentations of many TDI scenarios. When model is loaded horizontally FEA simulates buccal plate fracture. Understanding TDI biomechanics will allow better diagnosis, treatment, and prevention of dentoalveolar injuries.
ACKNOWLEDGEMENTS

I would like to thank the following people for their contributions:

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Dr. Jessica Lee, for her consistent open door, listening ear, and sage advice that she was willing to share.

My family, for always being my biggest fan and believing in limitless possibilities

The love of my life, my wife Jennifer, for her unwavering support, her constant love and devotion, and her awe inspiring effort as a wife and mother. She helps me want to be my best self. With her by my side no challenge is too great.

My daughter, sweet Adelaide Jayne, for her laughter and love. She is our reason for sacrifice and hard work.
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<table>
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<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>FEA</td>
<td>Finite element analysis</td>
</tr>
<tr>
<td>FEM</td>
<td>Finite element method</td>
</tr>
<tr>
<td>GPa</td>
<td>Gigapascal</td>
</tr>
<tr>
<td>MPa</td>
<td>Megapascal</td>
</tr>
<tr>
<td>PDL</td>
<td>Periodontal ligament</td>
</tr>
<tr>
<td>TDI</td>
<td>Traumatic dental injury</td>
</tr>
</tbody>
</table>
LIST OF SYMBOLS

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ξ Damping ratio

ρ Density

ν Poisson’s ratio

σ₁ 1st Principal Stress (tension)

σ₃ 3rd Principal Stress (compression)

E Young’s modulus
I. INTRODUCTION

Traumatic injury to the orofacial complex has been reported as a relatively common finding in population-based studies.\(^1\) The orofacial complex comprises 1 percent of the total body area while oral injuries account for 5 percent of total injuries with that percentage being increased in children.\(^2\) Reported prevalence of traumatic dental injuries (TDI) among school-aged children ranges from 15 to 30 percent.\(^1,3-8\) In the majority of cases, TDI involves the anterior teeth with the maxillary central incisors being most affected and mandibular central and maxillary lateral incisors less frequently involved.\(^9\) This trend is also seen in the primary dentition.\(^10\) Typically TDI affects a single tooth\(^11\), however sports and automobile accidents are predisposed to multiple tooth injury.\(^9\)

The epidemiology, etiology\(^12\), and treatment\(^13-17\) of traumatized maxillary central incisors have been well documented in the literature. Unfortunately minimal research has addressed the mechanical energy production and dissipation seen in dental injuries; the biomechanics of TDI is not well understood.\(^18\) There are limited *in vitro* dynamic/impact force studies to understand the mechanism of TDI.\(^19, 20\) TDI can result from direct or indirect trauma.\(^21\) Direct trauma is demonstrated by a blow to the tooth; typically in anterior teeth. Indirect trauma occurs when the lower jaw is forcefully closed against the upper jaw, by a blow to the chin, which may result in vertical fractures of the premolars and molars as well as fractures of the mandibular condyles or symphysis.\(^18\)
A number of factors describe/determine the destructive nature of a TDI, these include: 1) energy of impact, 2) resilience of the impacting object, 3) shape of the impacting object, and 4) direction of the impacting force. Energy introduced into the system is dependent on mass and velocity of the object which strikes the tooth. Theoretically, low velocity blows tend to cause more damage to the supporting structures, leaving tooth structure less disturbed. High velocity impact on the other hand would result in more tooth fractures with minimal disruption to the supporting tissues. Depending on the location and vector of the impact, the energy may be distributed through the tooth and into surrounding structures in specific patterns.

Previous TDI research has utilized *in vivo* animal models; however these studies are very expensive and may blur ethical lines. *In vivo* human trials which introduce traumatic force would clearly be unethical, while long-term prospective human studies would be very time consuming and very costly due to difficulties in following participants over many years. *In vitro* models are also limited due to the inability to accurately simulate important structural elements such as the periodontal ligament (PDL) and alveolar process. *Ex vivo* models would similarly lack integral soft tissues which are crucial to understanding impact propagation and dissipation. Many of the questions concerning impact propagation in permanent tooth TDI may be more accurately addressed using complex computer simulations known as finite element analysis (FEA) or finite element method (FEM). Finite element method as defined by Ko and co-workers is “a numerical approximation to solve partial differential equations (PDE) and integral equations that are formulated to describe physics of complex structures (like teeth and jaw joints).”
II. REVIEW OF LITERATURE

FEM has been used to study complex dental biomechanical questions. Endodontics\textsuperscript{25, 26}, restorative dentistry\textsuperscript{27-28}, and orthodontics have benefited from FEM. In orthodontics FEM has been used to investigate static treatment forces.\textsuperscript{24, 29-35} In TDI the resultant forces are not static or sustained but rather dynamic or ramped forces which are initiated and decay rapidly. FEM is used by mechanical engineers to safety test prototypes.\textsuperscript{36} Few 2-D\textsuperscript{37-39} and 3-D\textsuperscript{40, 41} dynamic FEAs have simulated TDI impact.

A. TWO-DIMENSIONAL DYNAMIC FEA:

Three studies used 2-D dynamic FEA to investigate TDI. These include two by Huang et al\textsuperscript{37, 38} and one by Miura and Maeda\textsuperscript{39}.

1. Impact Angle vs. Stress Distribution in a Maxillary Central Incisor

In 2005 Huang and colleagues simulated one of the first dynamically loaded finite element analyses to simulate dental trauma.\textsuperscript{37} The aim was to determine the relationship between impact angle and stress distribution in a maxillary central incisor. A 2-D FEM of a single maxillary central incisor was constructed and transient dynamic analysis (responses of structures under time-dependent loads) was completed. Transient dynamic analysis is used for evaluating structures that experience damping effects.\textsuperscript{37}
Materials have components of perfect elastic solids or viscous liquids. During impact of a viscous material, the strain energy may be converted to another form. This phenomenon is called damping. Damping factor is defined as the fraction of strain energy lost in one deformation cycle. Their 2-D FEM included: enamel, dentin, pulp, PDL, and alveolar bone. The PDL width (0.25 mm) was obtained from previous study.

Huang et al constructed their model using generally accepted material properties for the hard and soft tissues (Table 1) while damping factor was derived empirically and applied to the model. A sinusoidal impact protocol was used with a peak force of 800 N, rise time of 2ms, and total duration of 4ms. Forces were applied at one node on the facial crown in three vectors, F1) perpendicular to long axis, F2) 45° labial to incisal edge, and F3) vertical force parallel to long axis. Von Mises equivalent stress contours were measured.

Vertical impact had the highest stress while horizontal impact remained relatively low. Peak stress trailed maximum loading by 0.05 ms which may be explained by damping effect. Horizontal impact forces resulted in horizontal crown fractures while vertical impact forces produced three fracture patterns: tooth neck, oblique crown-root, and oblique root.

<table>
<thead>
<tr>
<th></th>
<th>Young’s Modulus (GPa)</th>
<th>Density (gm/cm³)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Enamel</td>
<td>77.90</td>
<td>3.0</td>
<td>0.33</td>
</tr>
<tr>
<td>Dentin</td>
<td>16.6</td>
<td>2.2</td>
<td>0.31</td>
</tr>
<tr>
<td>Pulp</td>
<td>0.00689</td>
<td>1.0</td>
<td>0.45</td>
</tr>
<tr>
<td>Periodontal ligament</td>
<td>0.05</td>
<td>1.1</td>
<td>0.45</td>
</tr>
<tr>
<td>Alveolar Bone</td>
<td>3.50</td>
<td>1.4</td>
<td>0.33</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>10.00</td>
<td>1.4</td>
<td>0.26</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>0.50</td>
<td>1.4</td>
<td>0.38</td>
</tr>
</tbody>
</table>

Table 1: Material properties used in FE model for Huang
2. Damping Ratio and Impact Propagation in TDI

A follow up study was performed by Huang et al.\textsuperscript{38} to determine the relationship between damping ratio and impact propagation. For a material that vibrates which experiences viscous damping, “the damping force $F$ is proportional to the velocity $v$ and can be defined as $F = -Cv$ where $C$ is the damping constant.”\textsuperscript{38} Damping ratio ($\xi$) is the ratio of the damping constant ($C$) over the critical damping constant ($C_c$) where critical damping is the relationship between oscillatory and nonoscillatory motion: $\xi = C / C_c$.\textsuperscript{38} Damping ratio was previously determined by reverse calculation; while this study determined it directly in human subjects. The average of all the damping ratios identified in human subjects was applied to the model to reflect the damping properties of PDL and pulp tissue.

The 2-D FEM was similar\textsuperscript{37} to the previous study with the exception of differing damping ratios and resonance frequencies which were obtained via human subjects. The same physical properties were used (Table 1). The same sinusoidal loading protocol was used at loading vector $F_2$ (45° labial to incisal edge of the maxillary central incisor). Stresses at the labial incisal edge were evaluated when the damping ratio applied to the model was 0.1-, 1-, 10-, and 50- fold of the measured value. Stresses experienced by a static load of 800 N were also analyzed in comparison to the dynamic load which was the same as in the previous study.\textsuperscript{37,38}

Stress contours were similar to the $F_2$ vector\textsuperscript{37} of the previous study. Static FEM stress was similar to the dynamic model at 2.05 ms (maximum stress). Average damping ratio was 0.146, significantly higher than metal (0.01)\textsuperscript{44} therefore beside enamel/dentin the tooth has various damped tissues. These damped tissues provide protection to teeth by
dispersing stress over a prolonged time as well as reducing the maximum stress experienced.\textsuperscript{38}

3. Dental Avulsion of a Maxillary Central Incisor

Miura and Maeda approached TDI from a different perspective than Huang et al. The aim of their study was to simulate dental avulsion of a maxillary central incisor looking at stress distribution over time. The 2-D dynamic FEM included: enamel, dentin, pulp, PDL, alveolar bone, and compact bone. The PDL was slightly narrower than in the Huang model\textsuperscript{37, 38} at 200\textmu m which had been documented previously.\textsuperscript{45, 46} Material properties (Table 2) used in the model were obtained from previous research\textsuperscript{34, 47-50} and differed from those reported by Huang\textsuperscript{37, 38} (Table 1). The model was designed with a limit of 1.4x the original PDL width in order for avulsion to occur which was determined previously.\textsuperscript{34, 45} When the displacement of the PDL during impact exceeded the limit specified the stress between the nodes connecting the PDL to the tooth became zero instantly and the tooth avulsed. The impact was ramped to 100 N over

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s Modulus (MPa)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Enamel</td>
<td>50,000</td>
<td>0.3</td>
</tr>
<tr>
<td>Dentin</td>
<td>18,600</td>
<td>0.31</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>11,500</td>
<td>0.33</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>431</td>
<td>0.3</td>
</tr>
<tr>
<td>Pulp</td>
<td>2</td>
<td>0.45</td>
</tr>
<tr>
<td>Gingiva</td>
<td>200</td>
<td>0.45</td>
</tr>
</tbody>
</table>

Table 2: Material properties of finite-element model for Miura\textsuperscript{39}
1.5ms and was applied at 90° labial to the incisal edge. Von Mises stress and displacement were observed over time.\textsuperscript{39}

Loading protocol differed between the two investigations.\textsuperscript{37-39} Most apparent is the difference in shape of the force delivery: sinusoidal\textsuperscript{37, 38} vs ramped\textsuperscript{39}. Huang was testing stress within the system that is experienced through impact while Miura and Maeda were simulating avulsion. Force magnitudes were drastically different between the two research groups. Tooth disruption or fracture requires intense force of short duration, whereas soft tissue trauma (luxation, avulsion) forces should be lower in magnitude over a longer duration. Magnitudes chosen are consistent while the duration is reversed according to study aims. Sinusoidal loading protocol has been favored in impact simulation of hip fractures due to fall as well as frontal head injuries.\textsuperscript{51, 52}

B. THREE-DIMENSIONAL DYNAMIC FEA:

There have been many static 3-D FEM studies\textsuperscript{24, 29-32}, but to date there are a few dynamic 3-D FEA.\textsuperscript{40, 41}

1. Horizontal and Vertical Loading of a Maxillary Central Incisor

A recent study by da Silva et al.\textsuperscript{40} simulated TDI of a single maxillary central incisor within supporting hard and soft tissues. The tooth was loaded with a ramped force of 2000N applied over 4 ms. The forces were applied at two different locations: F1) horizontal load acting at 90 degrees to the buccal surface of the midcrown, and F2) vertical load in the center of the incisal edge. The model included enamel, dentin, pulp, PDL, cortical bone, and alveolar bone. Physical properties used were consistent with
previous studies.\textsuperscript{37, 38} The PDL space had an average thickness of 200μm. The number of elements totaled 241,940.

Von Mises stress index was used to quantify stress distribution. With horizontal loading (F1) higher stress was seen on the crown and cervical areas of the model. Most of this stress appeared to concentrate on the enamel while little stress was seen in the pulp and PDL. Maximum stress experienced at the buccal surface (point of impact) ranged from 198-296 MPa while the lingual surface of the tooth experienced stresses of 395-494 MPa. The cervical areas experienced the most stress with peak stress of 889 MPa.\textsuperscript{40} The pulp and PDL experienced significantly less stress with highest concentration for pulp being experienced in the crown with most stress on the palatal area and PDL concentration being at the cervical area and dissipating towards the apex.

For vertical loading the stress concentrations were found at the buccal crown, neck and bone with peak stress at 311, 700, and 389 MPa respectively. The pulp and PDL similarly experienced less stress compared to hard tissues as was seen in horizontal loading; however more stress was experienced in these tissues compared to that experienced with horizontal loading.

2. Multi-tooth Trauma Observed in Sport Injury

The only multi-tooth 3-D FE model simulating dental trauma was constructed by Casas et al.\textsuperscript{41} This dynamic 3-D FEM was constructed to simulate multi-surface blunt object/baseball trauma. A pitching machine was used to drive a baseball (142 g at 14 m s\textsuperscript{-1}—amateur speed\textsuperscript{53}) at a cadaver skull. Instruments recorded the baseball velocity/acceleration, forces, moments, strains, and energy dissipation of the maxilla. The
majority of the damage occurs in the first 0.44 ms prior to head recoil\textsuperscript{54}; therefore the cadaver head was fixed. The 3-D FEM of the maxilla included four incisors and two buccal tooth segments (amorphous canine to second molar).

Results from the FEA were compared with the principle strains observed in both experimental and epidemiologic research.\textsuperscript{41} Model mesh accuracy was confirmed via a convergence test through process known as h-refinement.\textsuperscript{55} The difference in magnitude between the calculated (FEA) and measured (cadaver) principle strains ranged from 1.7-11.4\% (Table 3). The results show a correlation between experimental and calculated models. The results observed mimic typical TDI under certain conditions.

<table>
<thead>
<tr>
<th>Anatomical location within cadaver and model</th>
<th>Measured maximum principal strain in 2-D ($\varepsilon$)</th>
<th>Calculated maximum principle strain in 2-D ($\varepsilon$)</th>
<th>Percentage of measured value (%)</th>
<th>Relative error (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>R. canine fossa</td>
<td>0.004243</td>
<td>0.004374</td>
<td>103.1</td>
<td>3.1</td>
</tr>
<tr>
<td>Medial palate</td>
<td>0.002371</td>
<td>0.0021013</td>
<td>88.6</td>
<td>11.4</td>
</tr>
<tr>
<td>L. canine fossa</td>
<td>0.003819</td>
<td>0.0038828</td>
<td>101.7</td>
<td>1.7</td>
</tr>
</tbody>
</table>

\textit{Table 3: Relative differences between observed and calculated maximum principal strain values in two dimensions on the cortical surface of the human maxilla.}\textsuperscript{41}

\textbf{C. DEFICIENCIES IN CURRENT KNOWLEDGE}

Although the 2-D FEM studies were relatively simple compared to the multi-tooth 3-D FEM they were more complete in anatomic accuracy. The only 3-D single tooth FEM was very accurate but lacked completeness within a full maxilla and in the context of surrounding teeth. Casas identified “Incorporation of tissue properties for dental hard tissues and increased anatomic detail of teeth” and “suspension of teeth in a periodontal ligament,” as potential FEM improvements.\textsuperscript{41}
Creation of a more complete 3-D FEM incorporating a full maxilla with anatomic accuracy and physical integrity that can be tested using realistic multi-tooth impact loading is essential for understanding the biomechanics of TDI. This information could not otherwise be ethically obtained in a sample population. TDI is very costly in terms of psychological/emotional distress caused to patients and their families, significant restorative costs over time, and substantial time and energy spent.\(^1\) Data generated may allow clinicians to more accurately diagnose, treat, and prevent dentoalveolar injuries.

**D. SPECIFIC AIMS**

In order to better understand the biomechanics of traumatic dental injuries, the following specific aims will be executed:

1. Construct a 3-D FEM of a complete maxilla including a complete permanent dentition (excluding 3\(^{rd}\) molars) with teeth consisting of enamel, dentin, and pulp attached to an alveolus containing cortical and trabecular bone via a periodontal ligament.

2. Introduce a static point force (200N) to single maxillary central incisor (#8) as well as both maxillary central incisors (#s 8/9) in three orientations:
   - F1) Horizontal (90° to buccal surface of midcrown)
   - F2) Oblique (45° to F1)
   - F3) Vertical (middle of incisal edge)

3. Evaluate the resultant stress (1\(^{st}\) and 3\(^{rd}\) principal stresses, and Von Mises stress), strain (1\(^{st}\) principal elastic strain), and displacement in the constructed FEM.

5. Introduce a dynamic force with a planar object on the maxillary central incisors and adjacent teeth to simulate a fall and determine the relationship between the exposure variables (acceleration and orientation of impact, and time) and the outcome variables (stress distribution—principal and Von Mises—and displacement).
III. MATERIALS AND METHODS

Finite element analysis is fundamentally a complex set of mathematical calculations.\textsuperscript{24} Finite element analysis allows a structure to be modeled with “discrete-element mathematical representation by subdividing it into simpler geometric shapes or elements whose apices meet to form nodes.”\textsuperscript{56} FEM requires an intimate understanding of the physical properties of each component of the system. With traumatic dental injuries the physical properties of enamel, dentin, pulp tissue, periodontal ligament, cortical and trabecular alveolar bone, and soft tissue is essential. For an accurate FE model under dynamic loading, the following physical properties are necessary to define the physical nature of each component: density (\(\rho\)), Young’s modulus (\(E\)), and Poisson’s ratio (\(\nu\)). Under static loading conditions only Young’s modulus (\(E\)) and Poisson’s ratio (\(\nu\)) are required.

Density (\(\rho\)) is defined as mass per unit volume (g/cm\(^3\)). Young’s modulus (\(E\)) is defined as the ratio of stress over strain within the limits of elasticity. Young’s modulus of a material is used to predict the lengthening or shortening of a material under tension or compression respectively. Poisson’s ratio (\(\nu\)) is defined as the ratio of transverse strain to axial strain or the degree of contraction (perpendicular to the applied load) when a material is stretched (direction of the load).

Finite element analysis uses computer modeling to answer biomechanics questions; therefore, many of the methodological considerations for strong study design
do not apply. There is no sample selection protocol. In biological models a FEM can be generated using cone beam computed tomography (CBCT) data from a single source subject. The information gathered from this image serves as a scaffold for constructing a three-dimensional finite element model. There is no control group. FEM is quantitative with absolute data, there is no variation—a model run under the same parameters will always yield the same result. FEM validation may include an experimental comparison group as was seen previously. Even with experimental validation there are limitations because of material constraints and anatomical inaccuracies incorporated into these models. Many industries have utilized FEM in engineering protocols and its mathematical algorithms have been validated over the past thirty years. Clinical observation continuity with the model will perform the function of validation. FEM construction is detailed below.

A. FINITE ELEMENT MODEL CONSTRUCTION:

The source subject for the FEM was an 18 year old healthy female with Class I malocclusion and crowding. The CBCT was taken previously for diagnostic and treatment planning purposes for comprehensive orthodontics. Permission to use this image for research purposes was obtained from the source patient. Three dimensional computer aided design (CAD) data for teeth (enamel, dentin, pulp) and supporting structures (PDL and alveolar bone) of a complete maxilla were obtained via a multislice (10 μm resolution) Micro CT scan (Micro-CT40, Scanco Medical, Basserdorf, Switzerland). Once the three-dimensional data was acquired interpretation of structures was performed using material Hounsfield values with a visualization program (Insight
Snap ITK-SNAP, NIH R03 EB008200-01, Paul A. Yushkevich/ University of Pennsylvania, Philadelphia, PA, USA). After segmentation of structures, contours were smoothed and isocurves were created (Geomagic Research Triangle Park, NC, USA). Model integration and finalization were completed (Dassault Systems SolidWorks Corp, Concord, MA, USA). An IGES file format of the model was loaded into ANSYS 14.0 (Swanson Analysis Inc., Hutson, PA, USA) and ICEM software was used to generate the finite element model.

The proximal contacts between the lateral and central incisors were very difficult to separate. Significant effort was made to establish nodal separation. All shared nodes between central incisors and between lateral/central contacts were identified in three planes of space. Once identified these nodes were duplicated, renamed and assigned to the proper elements. The desired result of this process was to prevent elements corresponding to different teeth to share the same node at the contact point. This would allow the teeth to react to force individually rather than be affected en masse. Unfortunately, despite the effort to establish nodal separation, the contacts remained connected. Components of the FEM were assigned material properties according to those reported in the literature\(^{56-62}\) (Table 4). The periodontal ligament space was set at 200 μm. Boundary conditions were defined within the superior extent of the maxilla (Figure 1).

<table>
<thead>
<tr>
<th></th>
<th>Young's Modulus (GPa)</th>
<th>Density (gm/cm(^3))</th>
<th>Poisson’s ratio</th>
</tr>
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<td>0.33</td>
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<td>Dentin</td>
<td>18.3</td>
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<td>0.31</td>
</tr>
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<td>Pulp</td>
<td>2.07 x 10(^{-3})</td>
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<td>0.45</td>
</tr>
<tr>
<td>Periodontal ligament</td>
<td>68.90 x 10(^{-3})</td>
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<td>0.45</td>
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<td>Cortical Bone</td>
<td>10.00</td>
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<td>0.30</td>
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<tr>
<td>Cancellous Bone</td>
<td>0.25</td>
<td>1.4</td>
<td>0.30</td>
</tr>
</tbody>
</table>

Table 4: Material properties proposed for FEM\(^{56-62}\)
B. STATIC POINT FORCE APPLICATION

As precursor to loading the model dynamically, the model was loaded statically (continuous force) at three different orientations: F1) horizontal load (90° to buccal surface of midcrown), F2) oblique load (45° superiorly through the buccal surface of the midcrown and F1), and F3) vertical load (midpoint of incisal edge) (Figure 2). Single tooth (maxillary right central incisor), as well as both maxillary central incisors were loaded with static point-force loading with all three loading scenarios (horizontal, oblique, and vertical). Stress distribution (1st/3rd principal and Von Mises), strain (1st principal elastic) and displacement were generated and analyzed based on loading protocol (horizontal, oblique, or vertical), teeth involved in loading (single incisor vs both incisors), as well as tissue isolated (full model, bone only, PDL only, teeth only). Stress/strain were measured in megapascals (MPa), displacement in millimeters (mm).
C. DYNAMIC IMPACT SIMULATION

In order to simulate TDI from a fall, a concrete block was fabricated using ANSYS 14.0 (Swanson Analysis Inc., Hutson, PA, USA) (Figure 3). A sinusoidal impact protocol will be applied to the generated FEM of the complete model. Computation of the FEM will be run using transient dynamic finite element software (LS-DYNA 3D Livermore Software Technology Corp, Livermore, CA, USA). Previous research has estimated the impact force generated in the hip joint during a fall onto concrete at 12.6 kN.\textsuperscript{63, 64} Impact orientation will be specified at the following locations: F1) Midfacial point (mesiodistally and incisogingivally) of the clinical crown of the maxillary central incisor (s), oriented parallel to the occlusal plane F2) 45° superiorly oblique orientation through the same midfacial point as in F1, and F3) the midpoint (mesiodistally) of the incisal edge with vertical orientation (90° angulation with respect to F1). In addition to varying impact orientation, impact acceleration will also be varied. Stress distribution (1\textsuperscript{st} and 3\textsuperscript{rd} principal and Von Mises), strain (1\textsuperscript{st} principal elastic) and displacement will be...
generated in the model and analyzed over time to investigate the effect of changes in impact orientation and acceleration.

Figure 3: Dynamically loaded concrete block will simulate a fall
IV. RESULTS

The finalized FE model of a complete maxilla included 764,124 elements. The detail provided in this model surpasses the next most intricate model\textsuperscript{40} for simulating dental trauma by more than 500,000 elements. It also has the ability to evaluate multi-tooth trauma as well as traumatic sequelae to adjacent teeth and supporting structures caused by single-tooth trauma. There is one limitation with the FE model constructed. When the model was initially built it was difficult to separate the contacts between the incisors. Subsequent effort was made to separate the contacts however in the testing of the statically loaded model the contacts continue to act as a single connected unit. This does not detract from the understanding that can be gleaned from the model, however any conclusions based on the generated results must be carefully interpreted due to this underlying compromise.

The results obtained during finite element analysis can be expressed visually with color mapping depicting intensity of the parameter of interest as well as numerically showing the range for that same parameter. In the case of this model the parameters of interest included: stress, strain, and displacement. Stress components such as 1\textsuperscript{st} Principal stress $\sigma_1$ (tension), 3\textsuperscript{rd} Principal $\sigma_3$ (compression), and Von Mises stress are analyzed. Von Mises stress, $\sqrt{(\sigma_1 - \sigma_2) + (\sigma_1 - \sigma_3) + (\sigma_2 - \sigma_3)/3}$, is a calculated parameter which takes into account the combined effects of all the principal stresses (Figure 4). The 1\textsuperscript{st} Principal elastic strain was only used to show how the PDL stretched (Figure 5). Von
Mises stress distribution was not used to show response of PDL due to the small Young’s modulus of PDL which results in a very narrow range for Von Mises expression.

Figure 4: Stress parameters: Tension, Compression, and Von Mises Distribution

Figure 5: 1st Principal Elastic Strain of PDL
The independent variables analyzed to better understand TDI include: A) Number of teeth involved (Single—200N vs. Double loading—400N), B) Orientation of static load (F1-Horizontal, F2-Oblique, or F3-Vertical), and C) Structures evaluated (Full model, Teeth alone, Bone alone, PDL alone. The dependent variables (outcomes) evaluated were stress—1st and 3rd Principal stress and Von Mises stress distribution, strain—1st Principal Elastic Strain, and displacement. Results will be presented first numerically and then explained visually.

Numerical output for all the variables generated during static loading of the FE model shows results for single loading protocol (200N statically loaded onto the right maxillary central incisor) (Table 5) as well as for multi-tooth loading (400N total, 200N statically loaded onto each of the maxillary central incisors) (Table 6).

Table 5: Peak Stress/Strain and Displacement for static loading of maxillary right central incisor
All loading scenarios evaluated showed that maximum displacement and peak stress was the same for the teeth alone as it was for the full model. For this reason peak values for the full model have been omitted from the tables. Teeth alone showed the most displacement and peak stress/strain compared to bone and PDL by themselves (Figure 6). In general more stress was experienced by bone compared to PDL while more displacement was seen in PDL compared to bone. The only deviation to this trend was with oblique loading (both single and multi-tooth loading). With oblique loading stress magnitude continued to be greater for bone alone compared to PDL, however displacement was equivalent between these structures when evaluated in isolation.

Table 6: Peak Stress/Strain and Displacement for static loading of both maxillary central incisors
Horizontal loading showed the most displacement and peak stress/strain compared to oblique and vertical loading (Figure 7). This figure also demonstrates that horizontal loading resulted in more bone bending (tension) in the buccal plate region which can be clearly seen by the intense color mapping in this area which is largely absent in the oblique or vertical loads. When evaluating peak stress oblique loading was greater than vertical, however vertical loading experienced more displacement than oblique. As the vertical loading protocol was based on a true vertical orientation perpendicular to the occlusal plane, not along the long axis of the tooth, a moment to the force was generated resulting in less intrusive luxation and more displacement of the crown buccally. When the load was doubled, the stress/strain and displacement were roughly doubled (Figure 8).
Figure 7: 1st Principal stress (tension) of a double load at F1, F2, and F3

Figure 8: 3rd Principal stress (compression): Comparison of single vs. double loading at F1, F2, and F3
V. DISCUSSION

Dental trauma has been estimated to affect 15 to 30 percent of all school-aged children.\textsuperscript{1} According to one study approximately half of adolescents experience dental trauma by the time they graduate from high school.\textsuperscript{5} TDI comes at a high price, whether it be the psychological or emotional turmoil caused to patients and their families who experience dental trauma, or the shear cost in dollars and cents to restore patients back to health over a number of years, or even the amount of time lost from work and school to rehabilitate affected individuals.

Studies have found that dental trauma can “impair [a] child’s social functioning, emotional balance, and well-being.” \textsuperscript{1} Even when dental injuries are managed well, successful treatment appears to reduce but not eliminate negative psychosocial impacts.\textsuperscript{65} For those who are very young, dental trauma can be very frightening and unsettling, both patients and families understandably have concerns about the long-term prognosis and outcome of a TDI. The financial costs of TDI are difficult to determine and are impacted by many different variables such as the age of the patient at the time of injury, type and severity of dental injury, type of treatment to rehabilitate the patient and prognosis of that treatment, frequency of replacement of treatment provided, and current and projected dental fees.\textsuperscript{66} Costs associated with lost time and productivity can also be significant. Time associated with the treatment of patients who experience TDI has been estimated to range between 3 and 17 office visits to rehabilitate the patient.\textsuperscript{67}
It is imperative that clinicians have the tools necessary to accurately diagnose dental injuries including all hard and soft tissue injuries, provide long lasting treatment to restore esthetics and function which will resist future trauma, and ultimately to provide realistic strategies to prevent dental trauma. Until recently clinicians did not have adequate tools to understand the biomechanics of dental trauma. Animal models had provided greatly to our understanding of the physiology/biology of dental injuries, sequelae, and the body’s healing mechanisms. Moving forward, however, these models are very expensive and the generalizability and ethics of these models are compromised. In vitro and cadaveric models are unable to provide the whole story as the soft tissue supporting structures are absent. Finite element analysis has the ability to address some of the challenges previously encountered. FEM may provide information to better understanding TDI.

Most dental trauma is experienced in the maxilla with the central incisors being most affected.\textsuperscript{9} Although most TDI is experienced by a single tooth, it is naïve to consider that these injuries happen in isolation and there are many injuries that result in significant multi-tooth trauma.\textsuperscript{11} It is for this reason that fabrication of a complete maxilla with accompanying dentition and hard and soft supporting structures is crucial to our better understanding the biomechanics of TDI. Creation of a full arch model is novel, it has only been accomplished once previously for the purposes of simulating dental trauma.\textsuperscript{41} Other complex models have been developed for the purpose of simulating orthodontic movement under static loads.\textsuperscript{24, 29} The few numbers of complete maxillary models may be a byproduct of the complexity of building such a model. In our pursuit to build the most complete model possible to study TDI we too encountered formidable
odds. Significant computing power is necessary to utilize the power of a 3-D dynamically loaded model. Unfortunately at the present time our intricate simulated FEM model would require substantially more computing power than is currently available both physically and financially within the School of Dentistry. This does not detract from the overall success of the project but rather reminds us of the step by step nature through which development of an accurate computer model representing a complex biological model must follow. The dynamic chapter will be addressed in the future. Notwithstanding, many TDI trends were encountered as this model was statically loaded.

The force used for this model was either 200N for a single tooth loading or 400N total for a multi-tooth load. Again this is for a static loading protocol. Dynamically loaded FE models have ranged from 100-2000N.\textsuperscript{37-40} Horizontal, oblique, and vertical loading protocols have been applied to previous models with horizontal and vertical loading being more prevalent than oblique. Vertical loading was found to experience the most stress by Huang and coworkers\textsuperscript{37}, whereas horizontal loading was found to experience more stress/strain/displacement with our model. There may be several explanations for this discrepancy. The vertical loading protocol utilized by Huang et al was oriented at the midcrown, not the incisal edge. Their vertical load was introduced at the buccal surface of the midcrown, whereas our vertical loading protocol directs the force at the incisal edge.

When the vertical load orientation is altered to approximate Huang et al. (Figure 9), the displacement, 1\textsuperscript{st} Principal stress (tension) and Von Mises stress distribution are all increased, however 3\textsuperscript{rd} Principal stress (compression) decreases compared to vertical impact at the incisal edge. This change did increase stress/displacement; however the
values did not outpace those observed with horizontal loading. It is unclear why this is observed. Perhaps this could be explained by the intimately related nature of the anterior contacts found in this model. When the contacts are ignored as well as their improbable linking of adjacent teeth into the traumatic zone, the model can still generate a great deal about what is happening biomechanically to the tooth/teeth in question as well as their adjacent structures. The design of the model at present can be instructive for dental trauma to a patient with teeth splinted together (re-traumatization), or even a patient in late stages of orthodontic treatment.

The finding that most of the disruption to the dentoalveolar complex happens at a tooth-level is consistent with past research. Reduced disruption of the soft tissue supporting structure was consistent with da Silva et al who found that pulp and PDL
experienced the least amount of stress during dental injuries.\textsuperscript{40} The work of Huang and colleagues shed light on the damping effect of the PDL to distribute forces over time allowing the blow to be cushioned.\textsuperscript{37, 38} It was observed that PDL experienced more displacement with less stress than did bone alone. This seems to make sense clinically as PDL is a more aqueous structure allowing it to distort and is more compressible than bone. The only discrepancy with this trend occurred during oblique loading when displacement appeared to approach equivalence between bone and PDL. One explanation for this is that during oblique loading the crown approaches the palatal crestal bone with its horizontal component, however it then cannot be displaced further due to the vertical component that locks the tooth at that position, thus the PDL cannot be compressed further.

Even at 200N, which is near the low end of loading protocols suggested by other studies, there is significant stress generated in the system. Maximum compressive stress experienced by a single load can be upwards of $1 \times 10^6$ MPa. According to Huang\textsuperscript{37} enamel fracture may occur at stresses as low as 50 MPa. Therefore, although the model shows dramatic displacement it is unknown what displacement might look like when highly stressed areas fatigue and fracture. This is important as we seek to understand the circumstances that dictate a coronal fracture vs. luxation vs. a combined tooth/bone injury.

As has been addressed previously, a significant limitation of this study concerns the inter-related nature of the anterior contacts, making it difficult to parse realistic effects on directly impacted teeth/adjacent teeth and their supporting structures from distributed forces under splint-like behavior. This reality also makes it difficult to rely on
the quantitative output which may be more related to peak stress experienced at the contact and not at the impact site. This does not strip the model of usefulness in understanding TDI biomechanics; however interpretation and generalizability must be viewed through this lens of skepticism. Through that lens we see the predictive value inherent in the model to identify alveolar fracture of the buccal plate in horizontal loading. TDI may manifest as an alveolar process fracture (Figure 10). This model demonstrates the location where alveolar fracture is most likely to occur (Figure 11).

![Figure 10: Radiographic presentation of alveolar process fracture](image)

Although pulp and PDL are included in this model an additional barrier to this FEA accurately reflecting nature is the absence of gingiva and lip coverage. There is tremendous variety in the thickness of these tissues. Lip coverage especially could play an important role in better understanding TDI biomechanics. Research has shown that
children with increased overjet who are lip incompetent are at greater risk for dental trauma of the maxillary incisors. Exclusion of this soft tissue again makes generalizability difficult, however the results could be more illustrative of children with increased overjet and lip incompetence which is a group more predisposed to TDI.

Figure 11: Alveolar Fracture: A) FE model, B) Radiographic examination
VI. CONCLUSION

Dental trauma is an important problem fraught with diagnostic and restorative complexity; incredible cost to individually affected patients, their families, and society; and an area whose preventive strategies have not been well tested or understood. Current *in vivo, in vitro*, and population based studies are prohibitive to increasing our understanding of the process of traumatic dental injuries. Anecdotal evidence seen in case reports of TDI may shed some additional light but are not definitive. Finite element method provides an impressive alternative to quantitatively and realistically predict force propagation through hard and soft dentoalveolar tissues. This study produced the following contributions to the literature:

1. Developed an FE model of a complete maxilla representative of and 18 year old female
2. Created a concrete block that may be used to approximate TDI caused by a fall.
3. Statically loaded the model with horizontal, oblique, and vertical loads under single (maxillary right central incisor) or multi-tooth (both central incisors) loading protocols.
4. Teeth alone experience most stress/displacement compared to bone and PDL.
5. Horizontal loading results in more stress/strain/displacement compared to oblique or vertical loading. Fractures to the buccal plate are more likely in horizontal loading.

6. With this model when increasing the load by two fold, the stress, strain, and displacement are roughly doubled.

This research provides and important step in better understanding TDI biomechanics. This information is critical in refining clinical diagnostic practices, enhancing prognosis of restorative treatment in areas prone to trauma, inspiring TDI prevention best-practices, and informing better design and material selection for injury prevention.
VII. LITERATURE CITED


