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A validated finite element study of blunt trauma to the human maxilla

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A VALIDATED FINITE ELEMENT STUDY OF BLUNT TRAUMA TO THE HUMAN MAXILLA

by

Peter Krimbalis, B.Eng
Ryerson University, 2002.

A Thesis
Presented to Ryerson University
in Partial Fulfillment of the
Thesis Requirement for the Degree of
Master of Applied Science
in the School of Graduate Studies
(Department of Mechanical Engineering)

Toronto, Ontario, Canada, 2004

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Abstract

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Peter Krimbalis, B.Eng

A Thesis Presented to Ryerson University in Partial Fulfillment of the Thesis Requirement for the
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Six embalmed cadaver heads were obtained, prepared and subsequently impacted to the medial maxilla with a 142-gram baseball traveling at 14 m/s. Measurements of strain were obtained through the use of strain gauge rosettes located at the medial palate and both canine fossæ. Three dimensional finite element models of a dentate human maxilla were constructed for the purpose of investigating the mechanical response to a simulated blunt impact. Convergence testing revealed that a refined mesh with over 70,000 degrees of freedom was necessary to obtain sufficient accuracy within the analysis. The simulated load case involved a transient, dynamic impact to the medial maxilla with boundary conditions imposed at the buccal segments of the model analogous to the experimental case. Results were validated by a direct comparison to the displacements and principal strains gathered from experimental and epidemiological data. For the examined load case, displacements were highly localized at the anterior portion of the maxillary incisors. The comparison of experimental and calculated principal strains as a result of the simulated impacts revealed a 1.67 to 11.37% difference in magnitude.

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Chapter 1

1 INTRODUCTION

1.1 Introduction to Thesis

Blunt trauma to the face can produce devastating injuries to the teeth, soft tissues and surrounding bony structures that can require a lifetime of physical, cosmetic and psychological rehabilitation.

Approximately half of all children will experience some sort of dental trauma during childhood (Andreasen, 1994). Athletic activities represent some of the most serious injuries and produce 36% of all dental trauma (Bijur et al. 1995) and 10.4% of maxillofacial fractures. The severity of these injuries can range from a chipped tooth to fractures of the facial structures.

Oral rehabilitation may require highly specialized hospital services and can cost over \$25,000 for initial rehabilitation. The cost of such rehabilitation is only partially covered by insurance carriers and, as a consequence, oral-facial trauma often leads to litigation.

The study of blunt force trauma to the face began in the early 1960s with the use of cadavers in automobile crash-test simulations (Hampson, 1995). The main areas of facial impact included the mandible, zygoma, maxilla and the nasal complex. Since then, studies on the mechanical behaviour of the facial structures have been conducted with a large focus on the mandible, or lower jaw. The prominent location of the mandible leaves it exposed to injury. However, the most common site for oral-facial trauma is the maxillary incisors that represent 90% of all dental injuries (Bijur et al. 1995). The high incidence of injury to the maxilla and its accompanying *anterior* teeth provides the epidemiological evidence to support investigation of this area.

Although insights have been gained from cadaver testing, such as the impact tolerances of several bones, the general approach has usually been a high mass-low velocity impact with a large

contact area on the test specimen (Swearingen, 1965; Hodgson *et al.*, 1967; Nyquist *et. al*, 1986; Schneider & Nahum, 1974; Allsop *et al.*, 1988; Welbourne *et al.*, 1989). As a consequence, little work has been done on the study of low mass-high velocity impacts such as those produced by baseballs, hockey pucks or cricket balls that act upon a very concentrated region of the maxillofacial complex. This is surprising since it has been reported that, in general, facial fractures occur from direct trauma acting over a small surface area (Rupp, 2002).

An in-depth study of the mechanics behind concentrated, low mass-high velocity impacts to the maxilla could provide insights in understanding the effects of blunt force trauma on the maxilla and maxillary incisors. A robust, numerical approach such as the Finite Element (FE) Method could be used in conjunction with accurate representation of the geometry of the maxilla to create a dynamic model of blunt oral-facial trauma. The model could then be validated with data gathered in a concurrent clinical study of low mass-high velocity impacts to cadavers. Such a validated FE model would allow predictions to be made regarding the behaviour of the maxilla in other blunt impact scenarios and these could be compared with clinical observations.

1.2 Literature Review

1.2.1 Physiological Examination of the Maxilla

1.2.1.1 Anatomy of the Maxillofacial Complex

The fourteen bones of the face collectively represent the most complex array of bones in the human body. The main structures mostly occur in pairs and are: the two *nasal* bones, two *palatal* bones, two *superior maxillary* bones, two *inferior turbinate* bones, two *lachrymal* bones, two *malar* bones, *vomer* and *inferior maxillary* bone (Gray, 1918). The bones are supplemented by other structures that include the *mandible* and *zygoma* to form the complete structure of the face. Figure 1.1 shows some of the detailed structures of the maxillofacial region.

The principal function of these bones, specifically the mandible and maxilla, is to provide the mechanism for mastication. As a secondary and tertiary role, the facial bones provide avenues for the sensory organs and the respiratory and vocal organs (Gray, 1918).

The *superior maxillary* bones or *maxillae* create the entire upper jaw responsible for mastication. In addition, the bones of the maxilla also form the roof of the mouth (bony palate), and several smaller cavities, *fossae* and fissures.

The maxilla is comprised of four surfaces - *external (facial)*, *posterior (zygomatic)*, *superior (orbital)* and *internal (nasal)*. The *external* surface is by far the most visible with its lower portions extending as far *inferiorly* as the teeth. It displays several depressions or *fossae*, which correspond to the location of the *anterior* teeth, namely the *incisors* and the *canines*. These *fossae* are useful, particularly the more pronounced *canine fossa*, in that they serve as identifiable landmarks for examination and measurement of surface features. Figure 1.2 shows many of these surface features and their locations on the *external* or *facial* surface.

Figure 1.1: Roof, Floor and Lateral Wall of Nasal Cavity and Superior Maxilla (adapted from Gray, 1918).

Figure 1.2: Left Superior Maxillary Bone, Outer Surface (adapted from Gray, 1918).

The *internal* surface is most noted for the *palate process*, which serves as the horizontal division between the upper and lower portions of this surface. The lower portion creates a portion of the mouth, while the upper aids in the formation of the *nasal fossæ*. *Inferiorly*, the *palate process* forms the roof of the mouth with its familiar bumps and ridges and eventually articulates with the *inferior meatus* to create the *posterior palatine canal*. Another noticeable feature is the large open region located *superior* to the *internal* surface. This opening leads directly to the maxillary sinus otherwise known as the *Antrum of Highmore*. The overall pyramid shape of the maxillary sinus is bounded by the *posterior*, *superior* and *external* surfaces of the maxilla with bony walls so thin they are translucent.

The *posterior* (zygomatic) surface is separated from the *external* surface by a bony ridge. Its overall shape is convex and it is directed backward from the other surfaces (Gray, 1918). This

surface serves as housing for several dental blood vessels and nerves as well as the formation of the *posterior palatine canal* via articulation with the *bony palate*.

The *superior (orbital)* surface aids in the formation of the orbital floor. Its thin, triangular surface is smooth and is bounded by a number of structures including the *lachrymal notch*, the *bony palate* and the *anterior dental canal* that is responsible for the transmission of nerves and vessels to the *anterior* maxillary teeth (Gray, 1918).

The four surfaces previously described represent the main body of the *superior* maxilla. Along side these features there exists a number of processes-*nasal*, *malar*, *alveolar* and *palate*, which need to be identified. The *nasal process* is the most *superior* of the processes and articulates with both the *ethmoid* bone and the *frontal* bone of the skull. *Anteriorly*, the *nasal process* articulates with the *nasal* bone and *posteriorly*, aids in the formation of the *nasal duct*. The *malar process* forms components in both the *facial* and *zygomatic* surfaces and also serves as a location from which a portion of the *masseter* muscle arises (Gary 1918). The *alveolar process* is composed of mostly spongy *trabecular* bone and represents some of the thickest regions of bone on the maxillofacial structure. It mainly serves as a receptacle for the teeth and is punctuated with several deep cavities to accommodate for them. The shape of these cavities varies with tooth functionality; the broadest of which accepts the molars and the deepest for the canines. Aside from its tooth reception role, the *alveolar process* is also home to the *buccinator* muscle, which arises from its outer surface (Gray, 1918).

The *palate process* as previously mentioned is a horizontal projection associated with the *internal* surface of the *superior* maxilla's main body. Figure 1.3 shows the *inferior* surface of the *palate process* and some of its main surface features. Note how the concave *inferior* surface is littered with *foramina* that create passages for numerous blood vessels and nerves. Close inspection of the *inferior* surface reveals some very distinct features. First and foremost, immediately behind the incisors there exists the *anterior palatine fossa* which is a large, oval shaped hole formed by the articulation of the two *superior* maxillary bones (Gray, 1918). Several canals which house various nutrient vessels extend from the base of the *anterior palatine fossa* and reach as high as the *nasal fossæ*. Other notable features of the *inferior* surface include the *greater palatine foramen* and the *lesser palatine foramen* located at the *posterior* border of the *palate process*. The *superior* surface of the *palate process* presents itself with some very distinct features as well, most striking of which is the *anterior nasal spine*, a sharp process located at the base of the nose. Behind it the *superior* surface raises itself into a ridge called the *nasal crest* that enables the reception of the *vomer*. A more

detailed review of the anatomical features of the maxillofacial structure and other components of the human skull are beyond the scope of this thesis.

Figure 1.3: The Palate and the Alveolar Arch (adapted from Gray, 1918).

1.2.1.2 Anatomy of Permanent Dentition

A normal adult human mouth contains 32 permanent teeth. The arrangement of these teeth into a semi-elliptical shape gives rise to common use of the term “dental arch”. For the purposes of examination and charting of activity, both the lower and upper dental arches are divided in half giving rise to four distinct quadrants of the mouth. A typical quadrant contains (working from *anterior* to *posterior*) a central incisor, a *lateral* incisor, two premolars (also referred to as bicuspid) and three molars. The actual numbering of the teeth has two common conventions. The Palmer notation, which is used throughout the United States of America, dates back to 1876. The Fédération Dentaire Internationale (FDI) two-digit system is used throughout Canada and most of the rest of the world. In the FDI system, the four quadrants of the mouth are numbered according to their position such that 1

is designated to the upper right, 2 to the upper left, 3 to the lower left and 4 to the lower right. The teeth themselves are also numbered from 1 to 8, with 1's referring to incisors, 2's to *lateral* incisors, 3's to canines, 4's and 5's to the premolars and 6's, 7's and 8's correspond to the first, second, and third molars respectively. The actual location of each tooth is then identified as two-digit combination of quadrant and tooth. For instance, 11 refers to the upper right, central incisor and is read one-one not eleven. Table 1.1 shows the currently accepted view of a FDI notation chart with the patient's right appearing on the left as if the patient was facing the viewer. It is with this numbering system that each tooth's location is easily and readily found.

Upper Right Quadrant	Upper Left Quadrant
18 17 16 15 14 13 12 11	21 22 23 24 25 26 27 28
Lower Right Quadrant	Lower Left Quadrant
48 47 46 45 44 43 42 41	31 32 33 34 35 36 37 38

Table 1.1: FDI Notation Chart.

Furthermore, since each tooth has multiple surfaces associated with it, each surface is named so that specific portions of the tooth may be referred to precisely. The outer surface of the incisors is referred to as the *labial* surface since it is adjacent to the patient's lips or *labia*. The inner surface of the teeth has two names depending on whether it is referring to the upper or lower dental arch. For the lower arch it is termed the *lingual* surface due to its close proximity to the tongue, and for the upper arch it is called the *palatal* surface due to its association with the palate. The sides of the teeth facing the cheeks are called the *buccal* surfaces derived from the name *buccinator*, which refers to the main muscle forming the cheek. Finally, the broad flat surface on top of the premolars and molars is referred to as the *occlusal* surface and the sharp cutting surfaces on the tops of the incisors are called the *incisal* edges.

The tooth is a simple structure that consists of some of the most unique materials and structures found in the body. The shape and size of each tooth is governed by its function and location within the mouth. Figure 1.4 shows a longitudinal section of a typical tooth, its components and some of its surrounding structures. The easiest way to begin identifying the various portions of the tooth is to simply look at the gum line. All portions visible beyond the gum line are referred to as the *coronal* portion or simply the crown and all portions not visible are deemed to be part of the root. The crown is covered by a white, protective layer called the *enamel*, which due to its high mineral content, is by far the hardest structure found within the body. The enamel contains no nerve endings

and is thus not sensitive making it an ideal substance for the mechanical breakdown of food through mastication. The coronal portion protected by the enamel is called the anatomical crown and portions that are visible beyond the gum line are called the *clinical* portion.

Underneath the hard enamel layer exists the *dentin*, which is a yellowish-white bony substance responsible for the bulk and shape of the tooth. It is this layer of dentin that through a process of mineralization initiates the formation of the outer protective coating of enamel. Deep within the dentin layer exists the soft pulp of the tooth. It is here that all the nutrient vessels and nerves are supplied for the dentin. The pulp can be thought of as a precursor to dentin in that as an individual ages pulp is continually transforming into dentin. It contains the cell bodies of odontoblasts that secrete the protein matrix that is then calcified into dentin. By the time a person reaches middle age, the vast majority of their tooth is composed of dentin.

Figure 1.4: Longitudinal Section of a Tooth.

Hidden above the gum line is the anchoring or root portion of the tooth. A thin layer called the *cementum* covers the root's surface dentin. From the cementum a tough, fibrous membrane called the *periodontal ligament* arises and serves as the actual mechanism for holding the tooth in the *alveolar* bone's socket. Contrary to popular belief, there is no actual bone-to-bone connection of tooth to its socket; instead, it is the collective work of the cementum, periodontal ligament, and the alveolar

bone that creates the hammock-like structure that supports the tooth within the jaw. (Camp *et al.*, 2000).

At the base of the root there exists a small opening in the root called the *apical foramen* that allows for the delivery of nutrients to the tooth from the surrounding alveolar bone. In cases of acute periodontal disease it is through the *apical foramen* that the infection passes when spreading to the periodontal tissues (Dentistry and Real Life, 2003).

1.2.1.3 Fractures of the Maxillofacial Complex

Though not nearly as common as long bone fractures, the maxillofacial complex is also subject to its own bony fractures. The force required to induce a fracture in a facial bone is quite large (up to 200lbs for the maxilla; Schneider & Nahum, 1974) and as a consequence a large amount of severe, soft tissue damage tends to accompany most maxillofacial fractures.

The earliest descriptions of maxillary fractures began with Rene LeFort at the dawn of the twentieth century. According to legend, LeFort dropped skulls off of the roof of a tavern and then examined the subsequent fracture patterns (Richardson, 2000). From these fractures, LeFort identified three distinct regions where injury was most likely to occur. These regions are best described anatomically as the *transmaxillary plane*, the *pyramidal plane* and the *craniofacial plane*. Using these three planes as a guide, LeFort then classified the types of fractures he saw into three categories of increasing severity namely LeFort I, II and III. Clearly the most distinguishing feature of a LeFort fracture is the resulting floating fragment; a section of bone which has been completely severed from the remainder of the face. This floating fragment yields the telltale elongated face of a typical LeFort II or III fracture. Figures 1.5 and 1.6 show both *frontal* and *lateral* views of LeFort complex fractures.

Figure 1.5: Frontal View of LeFort Fractures (adapted from Richardson, 2000).

Figure 1.6: Lateral View of LeFort Fractures (adapted from Richardson, 2000).

A LeFort I fracture, also known as a *transmaxillary* fracture involves a transverse fracture above the maxillary floor and below the orbital floor. The fracture lies above the *root apices* and may or may not involve the *medial* and *lateral* walls of the *maxillary sinuses*. The fracture often transverses the nose and will invariably involve the *pterygoid process* of the *sphenoid* (Richardson, 2000). Athletic injuries rarely produce more than LeFort I fractures.

A LeFort II fracture exploits the weakness of the *subzygomatic* plane. This plane is commonly referred to as the *pyramidal* plane since it is really two planes in one that resemble a pyramid with its uppermost vertex lying just above the bridge of the nose. As one would expect, a typical LeFort II fracture transverses the nose, *infraorbital rim* and ultimately the orbital floor. The fracture can proceed as far *laterally* as the *lateral buttress* and as far *posteriorly* as the *pterygomaxillary buttress*.

A LeFort III fracture is by far the most severe of the classic LeFort fractures. It is also referred to as a *craniofacial disjunction* since it involves a complete disassociation between the face and the cranial base. An injury of this magnitude is very uncommon and requires tremendous forces present only in high velocity impacts such as motorcycle accidents. In general, LeFort injuries are not present on their own; rather, they exist in multiple combinations with other LeFort fractures and fractures of the limbs, skull and/or spinal column (Richardson, 2000).

While the idea of fracturing one's face is definitely one of the most basic and horrifying of fears, not to mention serious at any level of severity, there are very distinct injury patterns that vary greatly in their disruption of the many soft and hard tissues of the maxilla. It is this fear that lead to the investigation of the underlying mechanisms behind fracture of the face due to blunt trauma.

In 1962, Huelke began a series of investigations in order to understand the factors behind acute facial fractures. One of the larger concerns was the apparent large range of values for the tolerance of facial bones. Gadd et. al (1968) reported that the relative strength of female facial bones is significantly lower than that of males. Aside from gender, the age of the victim also alters the strength of the bone with an eighty year old showing as much as a 30% decrease in strength (Yamada and Evans, 1970). Also, from individual to individual there may be significant changes in the strength of the bone depending on the size, shape and density of the bone sample. As a result, the next decade produced several arguments as to which factor governed the overall ability of a facial bone to tolerate and absorb impact before fracturing. Hodgson (1967) suggested that the peak maximum load was the most important criterion for facial fracture while Nahum *et al.* (1968, 1976) argued that the overall pressure exerted on the bone dominated. Amidst these disputes over which was the dominating criterion, it became evident that the impact radius and the time duration of the impact are strongly correlated with overall amount of force tolerable before fracture occurs. A concentrated, peak load can only be withstood for a very short duration of time and this peak load drops sharply as the load time increases. Using metal cylinders ranging from 2 to 16 lbs as impactors, Hodgson (1967) showed that impacts to the face that lasted for less than 4 milliseconds (ms) produced impact tolerances as high as 1000 lbs. When the force profile of the impact extended beyond the 4 ms range the impact tolerances peaked at only 250 lbs. When increasing the impact radius from 1 in² to 5.2 in², a 250% increase in tolerance before fracture occurred. As a result, it was concluded that since the facial complex can be thought of as a collection of simpler structures, peak force is dominant for concentrated loads, however, if the load is distributed over a number of bones (i.e. a large impact radius), a change in the failure criterion occurs that makes peak pressure (stress) the primary factor (Hampson, 1995).

Following Hodgson's work, several studies were conducted to verify his findings and extend his research. Much emphasis was placed on determining facial tolerance in terms of maximum force rather than pressure due to its ease of measurement and for comparison to Hodgson's work.

In order to expand the understanding of impact tolerances, it was decided that specific factors and specific regions of the face would be studied in depth in order to better understand their roles in fracture. The factors considered included degree of injury, impulse time history, test subject variation and impactor size. It was decided that a relatively large impact radius with a low impulse rate be used in order to precisely ascertain peak loads before fracture. The regions of the face in question were the *frontal*, *zygomatic* and *parietal* bones. It was decided that soft tissue would be left intact in order

to preserve its damping properties. From these tests, tolerances of 900 lbs, 225 lbs, and 550 lbs for the *frontal*, *zygomatic* and *parietal* bones respectively were determined (Gadd *et al.*, 1968). Subsequent testing revealed that the maxilla and the nose were the areas of weakness in the face with tolerance value of 150 lbs and 77 lbs respectively (Schneider & Nahum, 1974; Nyquist *et al.*, 1986). These findings agree with epidemiological data, which show that the vast majority of all facial fractures occur in these two areas.

1.2.1.4 Dental Trauma

Dental trauma may involve the teeth, the lips, gums, tongue and even the surrounding jawbones. Since the mouth is home to many varieties of soft tissue, most dental trauma is accompanied by significant amounts of swelling and bleeding.

In general, the most common site of injury to the mouth is the teeth themselves (Bijur *et al.*, 1995). Usually, the injury will only involve a single tooth; however, it is not uncommon for a severe injury to fracture or displace several teeth and the surrounding structures. Severe dental trauma must always be considered as an emergency situation since it may or may not include obstruction of the airway. In terms of preserving the function and cosmetic appearance of the mouth, it is also vital to consider dental trauma as an emergency since the window of opportunity to save a tooth diminishes rapidly with increasing time.

Crown fractures are the most common type of injury to a tooth representing 75% of all injuries to the permanent dentition (Andreasen, 1994). Statistically, the maxillary *central* incisors are the most prone to fracture, followed by the maxillary *lateral* incisors.

In general, there are only two basic types of tooth fractures: those that involve the root, and those that don't. From these two basic categories, a number of conventions exist to describe the type and severity of trauma. Included in these conventions are the three most common, the Andreasen, the World Health Organization (WHO) and the Ellis conventions. The Ellis and Andreasen conventions are slightly more comprehensive than the WHO convention. For the Ellis convention there are six types of crown fractures in total but in practice, only the first three are significant.

Starting from the WHO convention, the simplest of all crown fractures is the fracture of the enamel of the tooth. Under the Andreasen and Ellis (Ellis Class I) classification this means that little or no dentin is involved and in fact the enamel itself is not completely fractured. The fracture appears as hair-like cracks in the tooth's enamel that may run horizontally or vertically. An injury of this type

does not endanger the tooth considerably, although a rough edge may irritate the surrounding lips and tongue if left untreated.

The next type fracture is the complete fracture of the crown without involving the pulp. Both Andreasen and Ellis (Class II) consider this a fracture without pulpal involvement however, the exposed *dentin tubules* may allow the ingress of bacteria that can infect the pulp if left exposed (Bastone *et al.* 2000).

Logically, the next classification of crown fractures would be ones that include the pulp of the tooth. Andreasen deems this a complicated crown fracture while Ellis labels this as a Class III. Here, the tooth is fractured such that the pulp of the tooth is fully exposed and the tooth will likely bleed. Aside from considerable pain, the consequences include the eventual death or *necrosis* of the dental pulp if left untreated. If that is the case, root canal therapy will prevent the loss of a permanent tooth.

Injuries to the tooth that are above the gum line invariably involve the root and as a consequence are much more severe in nature. The WHO classifies the simplest of root fractures as ones that do not expose the pulp. This convention is analogous to the simpler crown fractures. WHO diverges from the Andreasen convention, which classified root fractures by sub-dividing them into complicated and uncomplicated crown-root fractures as well as complicated root fractures. Only the uncomplicated root fractures do not involve the pulp of the tooth. Ellis classifies a root fracture as either any “traumatized tooth that becomes non-vital, with or without loss of crown structure” (Bastone *et al.* 2000) or more severely, a complete fracture of the root, again with or without the involvement of the crown.

Once a tooth’s root has been traumatized, secondary injuries tend to simultaneously occur that may be more severe. These injuries are of the displacement variety in which the tooth’s position is displaced from its natural orientation. Depending on the amount of displacement, the vital tissues of the tooth may become necrotic. A displacement injury does not necessarily mean that the tooth has fractured, in fact it is common for the tooth itself to remain completely intact and yet its position in the mouth is altered. Displacement injuries always include damage to the socket itself and the loosening or jamming of the tooth within its socket. Andreasen, being the most comprehensive of the conventions, classifies the damage done to a tooth and its socket without subsequent loosening of the tooth as a concussion. The WHO calls the loosening of the tooth within the socket a *luxation*. Andreasen subdivides luxation injuries into *sub-luxations*, *lateral luxations*, *intrusions*, *extrusions*

and *avulsions*. The Ellis classification does not include injury to the socket (*luxations* and *concussions*) within its convention.

If a tooth is displaced completely from its socket it is universally deemed an *avulsion*. A lost tooth can represent a medical emergency if the avulsed tooth becomes aspirated or lodged into soft tissue. The consequences of an avulsion are not simply cosmetic. Mastication becomes problematic because malocclusion may occur in the absence of the tooth.

1.2.1.5 Summary

The craniofacial skeleton, with its highly irregular and specialized structure, represents the most complex array of bones in the entire body. Although their primary function is to provide a means for the mastication of food, the craniofacial skeleton also houses the sensory and respiratory organs making it a crucial part of the anatomy.

Each of the facial bones is easily identifiable from one another because of distinct features and landmarks such as the various *foramina*, *fossa* and *processes*. The most distinctive of all features is the array of 32 teeth that occupy the mouth of an adult human. These teeth are arranged into two arches for the upper and lower jaws. For charting and identification purposes, the mouth is divided into four quadrants and each tooth is numbered according to its position within the quadrants as dictated by the (FDI) two-digit system. Not only is each tooth numbered and identified, but also since a tooth is a complex structure, each surface of the tooth is labeled according to its position and portion of the mouth it faces.

The composition of a tooth includes many materials, the most notable of which is the enamel, which is the hardest substance in the body. Although mostly inorganic, a tooth is in fact alive and has both a healthy blood supply and innervation. The inclusion of a blood and nutrient supply makes a tooth just as prone and vulnerable (if not more) to infection and trauma as any other part of the body.

Dental trauma is no less an emergency situation than any other type of trauma involving fracture or displacement. Severe risks include aspiration of an avulsed tooth, fractures of the maxillary alveolar bone or the intrusion of a tooth into the maxilla or maxillary sinus. Dental trauma is described by a set of classifications and conventions for identifying the types of fractures and injuries that occur in the mouth. Some of these conventions are more comprehensive than others, such as the Andreasen classification system, but in general, the World Health Organization (WHO) classification system of dental trauma is an international standard.

Fractures to the maxillofacial complex, although very severe, are not common due in part to the very large amount of force required to fracture the facial bones. Classifications like the LeFort convention allow clinicians to properly diagnose patients with oral-facial trauma.

Much research has been done on the study of maxillofacial fractures, especially in the field of impact simulations. Although most of the investigations have focused on high mass-low velocity impacts, many insights have been made about the impact tolerances of the facial bones and the types of injuries that accompany their fracture.

Subsequent studies have demonstrated that impact tolerances vary with gender, different age groups and even among individuals. However, aside from these variations, there exists strong correlations among several studies and therefore the impact tolerances of the facial bones have been well established.

1.2.2 The Maxilla as an Engineering Structure

1.2.2.1 Mechanical Properties of Bone Tissue

The surface of bone represents an anisotropic, composite material with a range of material properties. When examined macroscopically it can be divided into regions of hard, compact bone called *cortical* bone, and softer, less dense bone called *cancellous* bone. If examined microscopically, the composite nature of bone becomes very evident with distinct sub-units known as *Haversian* systems or simply *osteons*. Though bone is a living tissue not unlike muscle or skin, it is composed largely of inorganic material. Two-thirds of its weight and half of its volume is comprised of a material known as hydroxyapatite, whose chemical formula resembles the following $3Ca_3(PO_4)_2 \cdot Ca(OH)_2$ (Fung, 1993). The remaining one-third of the structure is a tough fibrous material named *collagen* whose fibres serve as the matrix on which the hydroxyapatite crystals are aligned.

The design of bone itself is optimized for strength and unlike other materials its strength is only very loosely correlated to its density (Fung, 1993). Instead of having a dense inner structure like that of a metal, bone transmits the load it experiences along filamentary flanges and welds making its density much lower. The use of a filamentary design concentrates the material along the load trajectory and thus makes the bone strong and dense only in the regions that will experience high levels of stress (Fung, 1993). The optimization principles behind the design of bone are known as *functional adaptation* and the *principle of maximum-minimum design*. The latter of the two is

exploited highly in the aerospace industry where it is vital to create a structure whose strength is maximized while its weight is kept to a minimum. Roux first described functional adaptation in 1895 where he stated that it was the “adaptation of an organ to its function by practicing the latter” (Roux, 1895). Essentially, this means that the design of a leg bone such as a femur or tibia is such because it is loaded in a particular fashion through standing and walking. It is the nature of the load itself that governs the design of the bone. This is a direct consequence of the fact that bone is a living tissue with nutrient and blood circulation. It is through this nutrient exchange that bone is either grown or resorbed by the body. In general, an increase in mechanical stresses will spark a growth in the bone making it stronger in order to compensate for this higher amount of stress. On the other hand a lack of mechanical stresses will lead to the resorption of bone tissue and thus it will weaken (anyone who has ever experienced a broken bone understands this concept first hand after the removal of his or her cast leaves the newly healed area atrophied and weak).

Although it may display a unique design and adaptive nature, bone is a material like any other and will invariably possess a set of elastic properties. In order to ascertain the precise values of these elastic properties much work has been done on the mechanical testing of bone through load deformation tests and ultrasonic testing. The variation between the values of density (ρ), Young's Modulus (E) and other properties varies greatly from bone to bone and person to person. However, there is agreement with regard to most of the physical properties of bone (Currey, 2002).

When measuring the mechanical properties of any material, certain characteristics of both the material and the load case must be kept in mind. For bone, these characteristics include whether or not the bone is wet or dry, the orientation of the load and the strain rate. Each one of these characteristics, on their own or in combination, can drastically skew the results of a mechanical test if they are not monitored.

In the natural conditions of the body, bone is in a wet state surrounded by the *periosteum*, which is a tough, fibrous membrane that separates muscle tissue from bone. It is not surprising then that a sample of dry bone displays very different elastic properties than an equal sized wet sample. Dry bone is much more brittle and will be stiffer along its longitudinal axis. As a consequence of this brittleness, dry bone tends to fail with very low amounts of strain and the site of fracture is lined with a sharp and clean break. On the other hand, wet bone covered with periosteum is more ductile and as a result will tend to produce a green-stick fracture, leaving stringy, delaminated edges.

It appears that different properties will be found as the orientation of the load case is altered. Experimental values reveal that this is in fact the case. The exact nature of this orientation

dependency is still uncertain although Reilly and Burstein observed a tendency towards a transversely isotropic continuum (Reilly & Burstein, 1975). More recent reports from Ashman *et al.* in 1984 suggest that this is not the case but nonetheless the values found in their study correlate strongly with others found in the literature. Table 1.2 reports (with some modifications by Currey 2002) some of the findings in Ashman's study of the mechanical behaviour of the human femur, data for this table.

Though not as prominent as the orientation effects, bone is also a material whose Young's modulus is somewhat strain rate dependent. The viscoelastic nature of bone is the cause of this strain rate dependence and work by Carter and Caler (1983) show that an increase in strain rate of a thousand times will yield a Young's modulus whose value is 40% higher (Currey, 2002). If this is the case, it is important that all mechanical testing of bone be conducted at low strain rates to ensure that the results are not skewed.

Property	Tension (GPa)	Compression (GPa)
E_1	12.8	12.0
E_2	15.6	13.4
E_3	20.1	20.0
G_{12}	4.7	4.5
G_{13}	5.7	5.6
G_{23}	6.7	6.2
ν_{12}	0.28	0.38
ν_{13}	0.29	0.22
ν_{23}	0.26	0.24
ν_{21}	0.37	0.42
ν_{31}	0.45	0.37
ν_{32}	0.34	0.35

Table 1.2: Elastic Properties of Cortical Bone Taken from Currey, 2002¹.

1.2.2.2 Experimental Measurement of Strain from Bone Surfaces

The experimental measurement of strain from bone surfaces done both *in vivo* and *in vitro* is valuable within the realm of biomechanics for numerous reasons. Whether it is to determine the

¹ The sub-scripts 1,2 and 3 refer to radial, circumferential and longitudinal values, relative to the long axis of the bone (Currey, 2002).

mechanical response of a biological structure or to find the strain levels that spark functional adaptation within the bone itself, the investigator has a myriad of strain measuring techniques available depending on the application itself (Szivek & Gharpuray, 2000).

Several of the experimental strain measurement techniques can be grouped together based upon the similarity of their underlying principles of operation. Only a select few experimental strain measurement techniques will be discussed here.

Coating methods essentially involve covering the specimen in question with a thin layer of a bondable substance and loading the specimen accordingly. The idea behind coating techniques is that the strain experienced by the specimen will be transferred to the coating through the coating-specimen interface with minimal losses and thus the behaviour of the coating will provide insight to the behaviour of the specimen itself (Korioth, 1992). How the coating reacts to the loading conditions is indigenous to the technique itself. Two common coating methods are the brittle-lacquer and the birefringent photoelastic techniques.

The brittle lacquer technique begins with spraying the surface of the specimen with a black coat of paint, allowing it to dry thoroughly and then followed by spraying a layer of strain-sensitive lacquer such as colophonium or stresscoat (Korioth, 1992). Once the layer has dried, and assuming that the coating has a much lower yield point than the specimen, a load is applied to the specimen and, as a consequence, to the lacquer as well. The lacquer will then crack as it is subjected to the load and distinct patterns will emerge on the coating. The load is then removed and an electrolyte is applied to the cracks. With the electrolyte still wet, an electrically charged white powder is blown over the crack pattern that adheres to the wet solution and leaves behind a stark, white on black contrast for examination. The result is a qualitative description of what has happened during the loading and although that insightful in its own right, it lacks any quantitative value, which unfortunately, limits its usefulness.

The birefringent photoelastic coating technique operates on a very elegant principle; essentially, there exists noncrystalline materials that are optically isotropic when in a state of zero stress that behave similarly to crystalline materials and become optically anisotropic when in a state of non-zero stress (Korioth, 1992). This change in optical characteristics is known as “temporary double refraction” and remains present for as long as the material is loaded (Dalley & Riley, 1991). If a specimen is covered with a doubly refractive material, loaded and light is shined upon it, a fringe pattern is produced from within the coating that is a function of the stress induced within the specimen. The fringe pattern can either be colourful or monochromatic depending on the type of light

used. Insight is gained by measuring the difference in angular phase shift that is linearly proportional to the difference in the principal stresses as experienced by the specimen (Dalley & Riley, 1991). The beauty of the birefringent photoelastic coating technique lies in its ability to give a complete distribution of strains on the surface of the specimen as opposed to regions of failure only. This gives the investigator a full view of all the stress gradients present and allows for valuable insights to be made in regards to the propagation of stress throughout the specimen.

As valuable as the previously mentioned techniques may be, both are limited to the realm of *in vitro* analysis of biological systems and are quite labor intensive to implement. Therefore, as a consequence, there is limited use of them.

Electrical resistance strain gauge application is by far the most common of the strain measurement techniques largely because of its ease of application either *in vitro* or *in vivo*. The direct measurement of strain from the surface of a specimen is an important feature of strain gauges. Their sensitivity, and resistance to corrosion are also attractive advantages. However, since strain gauges measure strain at discrete points only, investigators must have a firm understanding of the test specimen so that the gauges are mounted at critical locations of interest. Keeping that in mind, the use of strain gauges alongside analytical models and material properties, can provide valuable inferences about the stress state of the bone itself (Szivek & Gharpuray, 2000).

The concept of using an electrical resistance strain gauge is analogous to the coating methods previously mentioned in that if a gauge is properly bonded to the specimen, and the sample is subsequently loaded, assuming small losses at the gauge-specimen interface, the gauge (specifically the wires within the gauge) will perceive the same strain as the specimen.

Lord Kelvin first discovered the fundamental principle behind the use of electrical resistance strain gauging in 1856. While loading copper and iron wires in tension he noticed that when wire is stretched so that its length increases, its cross sectional area decreases and its electrical resistance increases (Szivek & Gharpuray, 2000). The resulting change in resistance is given by the following equation:

$$R = \frac{\rho L}{A} \quad (1)$$

where R is the electrical resistance in Ohms, ρ is the resistivity of the conductor (Ohms•cm), L is the conductor length in cm, and A is the cross sectional area of the conductor (cm²). This change in resistance will invariably result in a change in the voltage across the resistor proportional to the

current applied to the circuit. This is a direct consequence of Ohm's law and is shown in the following equation:

$$V = IR \quad (2)$$

The change in voltage can be used to infer the actual strain on the gauge and thus the specimen's surface. By definition, strain (ϵ) is a dimensionless unit that is expressed as a ratio between the change in an object's length and its original undeformed length. Since strain is usually quite small, it is often expressed in terms of microstrain ($\mu\epsilon$), which is simply 1×10^{-6} units of strain (mm/mm for example).

In vitro strain gauging has been used since the 1950s, and this is where the procedure of tissue removal and chemical etching of the bone surface were first introduced. This procedure has proven valuable for the good bonding of gauges for *in vitro* studies, however, it serves to be far too aggressive for *in vivo* studies and modified versions of the technique must be employed (Szivek & Gharpuray, 2000).

In 1966, Roberts outlined several techniques to apply gauges to living animals. Some of the more poignant suggestions he made include the use of a drying, degreasing agent such as anhydrous acetone, ether or isopropyl alcohol and the use of instant setting methyl-2-cyanoacrylates instead of a dental epoxy (Szivek & Gharpuray, 2000). Included in the report was the first use of Teflon® (E.I. du Pont de Nemours and Company, Inc., Wilmington, DE) coated wires which insured insulation against the wet environment of a biological system. Roberts' suggestions have become widely accepted and are currently used for both *in vitro* and *in vivo* studies, largely because of their minimal tissue disruption.

Foil gauges similar to the one shown in Figure 1.7 first appeared in 1972. Foil design allowed for much easier bonding of the gauge to the bone surface and thus insured a minimal amount of disruption of the surrounding tissues. The advent of the foil gauge and its ease of application allowed for the measurement of strain from bone surfaces during normal physiological activity such as gait. This greatly expanded the knowledge base and the use of foil gauges to analyze complex force systems in the body continues today.

At the present time, *in vivo* studies of bone strain is still highly impractical on humans, especially in the complex and delicate structures of the maxilla. However, work has been done *in vitro* on humans beginning with Mongini *et al.* in 1981 and continuing on with Koriath in 1992. Both studies have focused on the mandible and were loaded such that a reproduction of the forces caused by the *masseter*, *medial pterygoid* and *temporal* muscles was achieved (Koriath, 1992). Although a

direct comparison to other studies was difficult to make due to the locations of the gauges (placed *bilaterally* at the mandibular angular regions), insights were made into the nature of the force system surrounding the mandible. At the time of this review, no work has been conducted regarding the experimental measurement of strain on the *superior* maxilla using electrical resistance strain gauges.

Figure 1.7: A Typical Foil Electrical Resistance Strain Gauge (adapted from Szivek & Gharpuray, 2000).

As previously mentioned, in order to fully exploit the benefits of using electrical resistance strain gauges, the investigator must fully understand the nature of the problem and customize the type of gauge assembly to be used in order to optimize the gauge type for the environment at hand. There are a myriad of factors behind the judicious selection of a gauge and it is a combination of these factors that will determine the overall effectiveness of the gauge assembly. Since the focus of this study is a short term *in vitro* experiment, details regarding the precise gauge selection process for *in vivo* studies will not be included.

The first of these factors is the foil material. There are three main tasks that must be satisfied by a foil. These include low temperature sensitivity, high strain sensitivity, and electrical resistivity. Most commonly available gauges are made of a nickel-copper alloy or a nickel-chromium alloy.

Temperature sensitivity of a gauge is important because it can lead to errors in measurement due to the change in resistivity it causes. In the event of a drastic temperature change after gauge installation, a gauge will register a strain even though one may not exist. Though this change is

negligible at room temperature, it will be a factor in high or low temperature scenarios. A properly selected gauge backing will aid in the heat dissipation/absorption process, and thus reduce this error. Many foil materials display good thermal stability and linear strain sensitivity over a large range of strain (Dalley & Riley, 1991).

Gauge resistance is also a key factor. A gauge is simply a resistor and analogous to a toaster oven or a space heater, a resistor will invariably increase in temperature as it resists the current passing through it. The highest resistance gauges produce the least amount of heat, and also aid in heat dissipation. Due to bone's poor thermal conductivity, a high resistance gauge should be used. A minimum of 120 Ω gauge should be used, although 350 Ω is recommended (Szivek & Gharpuray, 2000).

Much like the coating methods, a strain gauge will only perform well if it is securely bonded to the surface from which it is intended to measure. This means that any sharp discontinuities as a result of the geometry of the specimen must also be included in the geometry of the gauge. As a result, another key factor in the judicial selection of a gauge assembly is the type of backing material that houses the foil. The backing material provides a means of handling the gauge, bonding the gauge and as an insulator between the specimen and the foil itself. The most common backing material is polyimide (Szivek & Gharpuray, 2000). It corresponds to the backing material of choice for bone applications because of its flexibility and toughness. Since most bone surfaces are substantially curved (especially in the maxillary region), a flexible backing material is crucial.

A strain gauge measures the average strain over its own surface area. Since some specimens may experience high strains and high strain gradients, it is important that the narrowest grid width be chosen. Since bone behaves anisotropically, a narrow gauge will also help minimize inaccuracy along its width. However, a short gauge provides poor heat dissipation, and tends to be difficult to handle. One must decide which factors play the largest role in the specimen's analysis. In the case of an *in vitro* study of the cadaver maxilla, small gauges are of prime importance due to the large curvature of the skull's features, and the low temperature environment.

Gauges also come in a variety of patterns, as well as single versus multiple grids. Single grids are mainly applicable to situations where the strain acts along one direction primarily. Rosette gauges, which contain three or more elemental gauges, provide the most complete information about the strain state of a single point on the bone (Szivek & Gharpuray, 2000). They are useful for finding principle strains as well as strains along a specific direction. Two styles of rosettes exist: single plane and stacked. A single plane rosette implies just that, all the gauges lie within the same plane while a

stacked rosette has three (or more) gauges stacked on top of a single point. The single plane rosettes provide better heat dissipation, however their use is limited to very flat specimens. Principal strain measurements can be inaccurate when using single plane gauges on a curved surface with large strain gradients since each gauge collects data from a slightly different location (Szivek & Gharpuray, 2000). The stacked rosette measures strains in different directions at a single point, and is much more suitable in areas with high strain gradients and/or curved surfaces with an uneven or small radius of curvature (Szivek & Gharpuray, 2000).

Important to note is the fact that misalignment errors are less critical when employing rosette style gauges if the aim is to measure peak or principle strains. However, proper gauge alignment should not be abandoned because of this fact. It simply implies that in the situation of complex surface geometry of the specimen in which it is difficult to find specific axes or landmarks, the use of a rosette gauge will still allow for the measurement of peak strains with less dependence on their actual alignment versus a non-rosette style gauge (Perry, 1969).

In biological applications, it is common for the gauges to contain encapsulated lead wires and terminals meaning that the solder tabs, strain relief and wires are all contained within the backing material. This allows for greater waterproofing of the gauge, as well as taking up less space on the specimen if area is a concern. As a beneficial side effect, it allows for a direct connection to a signal conditioning and amplification system without the use of separately bonded and soldered terminals.

Although much easier than other experimental strain measuring techniques, electrical resistance strain gauging is not without its own set of laborious procedures and preparation. For the case of short-term *in vitro* experimentation, the prevention of fluid infiltration to the gauge-wire junction is not as important as that for long-term *in vivo* testing. The gauges can be sufficiently protected by purchasing gauges with the attached lead wires fully encapsulated. Additional coating of nitrile rubber may be added if a 24-hour curing time is available, however, this is not necessary for a short-term test where liquid ingress is not a problem.

Though the standard protocol of gauge bonding does apply to the case of bone, a slight modification of the process must be employed to ensure proper attachment of the gauge. (For a full description of strain gauge attachment and surface preparation techniques employed in this study please see the Experimental Methodology section)

The use of cyanoacrylates as a bonding agent is common for bone specimens. In particular a methyl-2-cyanoacrylate is useful due to its rapid polymerization and strength. Before any bonding can occur, the surface must be properly prepared. This involves the removal of the soft tissue and

periosteum. The location of the gauges should be clearly etched, or burnished onto the bone and then thoroughly dried. Wright and Hayes (1979) suggested a thorough degreasing and aggressive abrasion of the surface be carried out prior to gauge adhesion; however, this is not absolutely necessary for satisfactory bonding of the gauge (Szivek & Gharpuray, 2000). As a precautionary measure though, it is the author's recommendation that some degreasing and abrading take place in order to ensure that good contact and adhesion occurs.

If the gauge is to be placed on an area of thin bone, or high blood vessel density of the surrounding tissues, the location should be sealed prior to the application of the adhesive. The first step in the procedure for sealing the area is the application of the adhesive catalyst to the bone surface and allowing it to thoroughly dry for at least five minutes. This is followed by the application of one to two drops of the bonding agent, spread thinly and uniformly over the testing surface. This adhesive is then allowed to thoroughly dry (at least 5 minutes) and it is then abraded lightly with 400 grit silicon-carbide paper.

After the gauges have been properly attached, it is important to develop an adequate signal conditioning and recording system. The signal conditioner must first balance the initial resistance of the gauge, wires and connectors. This is accomplished by the use of a Wheatstone bridge. The style of bridge (1/4, 1/2, or full) depends on the gauge configuration. For the case of a single gauge per channel, a 1/4 bridge is sufficient for balancing the system.

A recording system is then attached to the signal conditioner and must be capable of recording the voltage changes from the strain gauges accurately. The output of a gauge is in small (in the range of mV's) and the recording system must cater to this. A continuous system with multi-channel capabilities, a broad frequency response and a high slew rate is recommended. For the vast majority of *in vitro* experiments, a sampling rate of at least 1000 Hz should be selected.

The accuracy of the system can be checked against measurements done on specimens made of well-known materials and of simple geometries whose exact strain values can be determined via beam theory.

Finally, it is also a good practice to establish a point of zero strain that will be the benchmark for all other data points. The selection of this point of zero strain is arbitrary, and done on a case-by-case basis.

1.2.2.3 The Finite Element (FE) Method

The Finite element (FE) method is a numerical procedure for solving complex field problems. By definition, a field problem is one that requires the determination of the spatial distribution of one or more dependant variables that describe the physics of a problem via a complex system of differential equations or integral expressions. The underlying philosophy for the solution of a field problem via the FE method is that the complex problem can be divided into smaller domains (elements) via a process known as discretization. During the discretization, field variables are interpolated across these elements using polynomial expressions generating a set of algebraic equations, which upon summation can be used to formulate a global set of algebraic equations that can be solved via matrix operations (Reddy, 1993).

The collection of elements that make up the global system is known as the mesh. It is interconnected at specific points along the domain boundaries called nodes with specifically defined degrees of freedom known as boundary conditions. The algebraic solution of the FE mesh yields nodal unknowns (and its first derivative if the element is of higher order) that once combined with the assumed field in any given element, will completely describe the spatial variation of the field element within that element (Cook *et al.* 1989). The solution to a field problem by the FE method is analogous to building a brick wall one brick at a time. The solution to each element is the act of laying down a brick and once all of the bricks have been laid down (elements solved) the wall is complete and thus so is the solution for the global system.

While the FE method is a numerical technique, convergence to the actual solution is guaranteed so long as the physics of the problem is not violated during the discretization process. In order to maintain the integrity of the problem, elements must be both complete and conforming. An element, which is conforming and complete, will have as its interpolation functions polynomials that contain constant terms for the field variable and its first derivative.

The advantages of the FE method are numerous and substantial. Firstly, it is applicable to any field variable problem. A common misconception regarding the FE method is that it is limited to the realm of structural mechanics. This is simply not the case. Fluid flow, heat exchange and magnetic fields are all handled equally as well by the FE method.

Perhaps the single greatest advantage of the FE method is that during the discretization process the boundary conditions and the material properties for the problem/element domains are not required *a priori*. In fact, these properties may change from one element to another or even within the element themselves and are not specified until the final assembly process of the element domains.

(Cook *et al.* 1989). Most other numerical techniques require the specification of the boundary conditions *a priori* and thus should any changes to the boundary conditions be made, a complete reformulation of the numerical approximation to the continuum must be made.

The true power of the FE method lies in its ability to converge to a solution a number of ways, the simplest of which is to discretize the global domain into more and more elements which in turn allows for the interpolation of the field variable to operate over a smaller domain and thus approximate the solution closer to the actual continuum. A second technique used to increase the rate of convergence is the use of higher order elements. Here, the interpolation function used across the element domain is of higher order and thus can capture sudden changes in the spatial variation of the field variable more accurately than a lower order interpolation function would.

Applications of the FE method began in the 1940's with the pioneering work of Hrennikof and McHenry (Hrennikof, 1941; McHenry, 1943). Their use of a one-dimensional lattice of bar and beam elements in order to formulate the approximate solution of a continuous solid went largely unnoticed by the scientific community. A few years after Hrennikof, Courant proposed the use of variational methods such as the Rayleigh-Ritz method to calculate the solution to stress analysis problems (Courant, 1943; Logan, 1993). Courant also spearheaded the concept of using interpolation functions to obtain approximate numerical solutions. At this time however, high speed digital computers did not exist and thus the manipulation of Courant's cumbersome equations were difficult if not impossible.

In the early 1950's, Argyris and Kelsey first introduced energy principles into the FE method, which proved vital to the development of the FE method since the concept of positive energy required to do a non-zero displacement carries with it several important consequences (Argyris & Kelsey, 1954).

Up until 1956, all elements used in the FE method were one-dimensional. This was changed by the work of Turner *et al.*, who not only developed two-dimensional triangular (the simplest of all 2D elements) and rectangular elements, but they also proposed the concept of the direct stiffness method (Turner *et al.*, 1956). In the direct stiffness method, the FE method is able to obtain the complete stiffness matrix for the entire structure. This idea, along with a rapidly developing world of high-speed digital computers sparked a boom in the development of the FE method.

The sixties gave birth to the first three-dimensional elements with the development of the tetrahedral stiffness matrix by Martin and the axisymmetric solid by Clough *et al.* in 1965 (Martin, 1961). Work was also being done on large displacement, inelastic behaviour and thermal problems

that greatly opened the field of the FE method (Logan, 1993). Non-linearities were considered in 1962, and the extension of the FE method to any field problem application was proposed by Zienkiewicz and Cheung in 1965 and carried out to fluid flow and heat conduction by Wilson and Nickel in 1966 and Martin in 1968 (Logan, 1993).

In recent years, the finite element method has extended into non-linear, and large displacement dynamic systems, and many new and improved numerical techniques have developed to help solve such problems.

The FE method also has its place in the realm of biomechanics as well, since it enables the researcher to model the complex geometries with ease and accuracy. The inclusion of various material properties and nonlinearities allows the investigation of almost any biological system. FE analysis of biological structures like bone began in earnest in the 1970's. Most early work involved the structural analysis of long bones such as the tibia and femur for examination of the loads and strains encountered during normal gait. Much work has been done on the mandible in both two and three-dimensions by Koriath in 1992 as well as Hart *et al.* in 1992. Both of the aforementioned studies focused on fully and partially edentulous mandibles as well as the force systems created by the surrounding masticatory muscles. Vollmer *et al.* in 2000 conducted extensive work and created a validated model of the mandible. Investigating the mechanical behaviour of the mandible during simulated loads via electrical resistance strain gauging validated their model, which consisted of 7100 brick elements (Vollmer *et al.*, 2000). The results of their study showed a 99.2% correlation between experiment and model giving rise to the notion that complex biological system could be very accurately modeled and this model could be used as a predictive tool for further investigations (Vollmer *et al.*, 2000).

Though work has been conducted on the craniofacial skeleton (Tanne *et al.* 1987, 1991; Krabbel & Appel 1995; Voo *et al.* 1996; Sujith, 2000; Horgan & Gilchrist, 2003) most work has focused on the brain cavity, the cranium and the entire skull itself as it behaves in impact scenarios such as an automobile crash. No finite element modeling has been published on the maxillofacial complex in the region of the *superior* maxilla and the maxillary incisors. In general, most models of the craniofacial skeleton (and the maxillofacial complex) have been drastically oversimplified through the use of simple shell elements and compromised geometry. While that may have allowed comparisons between studies to provide insight into the behaviour of the skull as a whole, it does nothing to enlighten either researcher or clinician as to the mechanical and elastic behaviour of the maxilla as a solid, geometrically accurate model.

For a more comprehensive review of the FE method the following references are provided: Bathe, 1996; Becker, 1981; Huebner, 1975; Shames, 1985; Stang, 1973 and Zankiewicz & Taylor, 1989.

1.2.2.4 Summary

Bone presents a complex set of material properties accompanying its anisotropic, composite structure. Similar to teeth, bone is largely inorganic with a full two-thirds of its weight occupied by a crystalline substance called hydroxyapatite. Nevertheless bone is a living structure and has a generous blood and nutrient supply feeding its many substructures.

One of the most unique features of bone is its highly optimized structure, which boasts a trajectorial design. Stemming from the design principles of functional adaptation and the principle of maximum-minimum design, the trajectorial nature of bone is optimized such that the maximum amount of material is placed in regions of maximum force and stress.

Bone hosts a set of elastic properties that can be measured through experimentation. Other quantities such as experimental strain can be measured from a bone's surface and knowledge of these quantities will prove insightful to both researcher and clinician. Several experimental strain measurement techniques are available, such as brittle coating and birefringent photoelasticity. Each of these techniques is powerful and effective in its own right; none, however, is as common or robust as the electrical resistance strain gauging technique. Its ease of application and accuracy has made it an international standard in the field of experimental strain measurement.

Much work has been conducted on the experimental measurement of strain using electrical resistance strain gauges both *in vivo* and *in vitro*. From these tests, comprehensive guidelines and procedures have been developed to ensure good practice in both the judicious selection of an appropriate gauge assembly and its subsequent application to a specimen, be it *in vivo* or *in vitro*.

Clinical experiments are used to validate theoretical models of complex stress analysis problems. One of the most powerful techniques available is the Finite Element (FE) method. The FE method, simply put, allows the researcher to model the intricate geometries of complex structures and subsequently allows the quantification of the mechanical behaviour of the system in response to known loading and boundary conditions.

The FE method has been used with great success in past studies of biomechanical systems and has shown a strong correlation to experimental results when validated. This fact makes the FE

method an ideal choice for the examination of the mechanical behaviour of the maxilla during a low mass-high velocity impact.

Chapter 2

2 STATEMENT OF PROBLEM AND SCOPE OF STUDY

In order to fully understand the biomechanical behaviour of a low mass-high velocity impact to the maxilla, an experimental study of this precise scenario should be carried out. In this study, several variables must be constrained and further assumptions made in order to reproduce the impact in a manner that could serve to be useful in validating an FE model. In an ideal situation, this experimental study should be carried out *in vivo* but for obvious ethical reasons this is not possible and thus an *in vitro* examination of the impact must suffice.

A quantification of the strains experienced by the maxilla upon impact from an impactor moving at a known velocity will serve as the basis of a set of data, along with the overall reaction forces and moments of the impact to develop a working hypothesis that could explain the relationship between the controlled variables and the resulting deformation/fractures of the biological specimen. As with any experiment, particularly one involving a biological system, the simultaneous measurement of several variables (strain, force and moment) at multiple locations on the sample coupled with the very short response time of an impact presents a challenge. Although the data set will be somewhat incomplete due to the necessary assumptions in conjunction with the subsequent modeling of the system, the information gleaned from this analysis will prove useful in understanding the mechanical response of the maxilla upon impact.

If examined as a mechanical system involved in a transient dynamic load, a low mass-high velocity impact to the maxilla should be investigated as a system of two interacting bodies contacting over a distributed impact radius with the mass of one body being significantly lower than the other. The impacting body should be constrained such that the energy of the impact is absorbed wholly by the maxilla that should be rigidly constrained in order to do so. The location of the impact must also

be precise in order to avoid the distribution of the impact to other bony features of the craniofacial skeleton.

The FE method offers a method of modeling this impact while simultaneously preserving the integrity of the complex geometry involved in the experiment. It is necessary to validate the effectiveness of the FE method in modeling such an impact should it be used as a predictive tool for the quantification of the strains experienced by the maxilla during an impact. Therefore, it is the aim of this study to first experimentally recreate an impact scenario that is controlled and subsequently create, from biological CT data, a geometrically accurate model of the same impact that may be validated via a convergence to the gathered data.

Chapter 3

3 EXPERIMENTAL METHODOLOGY

3.1 Preliminary Experimentation and Preparation

In order to formulate a proper methodology and gain experience in the use of electrical-resistance strain gauging as a means of experimental measurement of strain, a series of preliminary experiments and exercises were carried out. The first test involved the mounting and preparation of gauges onto a highly oxidized steel beam in order to develop proper surface preparation procedures and establish the viability of the electronics. The second exercise involved surface preparation and mounting of strain gauge assemblies to a series of embalmed bovine ribs. This confirmed that the surface preparation and bonding protocol established earlier could be extended effectively to a biological surface with significant curvature and geometric complexities present. Finally, a mock representation of the actual experiment was carried out with full instrumentation on a plastic model of the human skull in order to verify all established protocols and the data acquisition system.

3.1.1 Application of a Strain Gauge Assembly to an Oxidized Steel Beam

The proper application of a strain gauge assembly involves a very strict protocol for both gauge and surface preparation in order to ensure the proper bonding of the instruments. In order to gain confidence in the application of strain gauge assemblies, an oxidized steel beam was chosen to properly prepare, bond and wire a series of strain gauge assemblies.

The surface of the beam was first degreased using isopropyl alcohol (other degreasers include acetone and ether) and wiped with sterilized cotton gauze. The procedure for wiping involved a single swipe in one direction only; any subsequent swipes required completely new sterilized cotton gauze. This procedure ensures that any contaminants removed are not reapplied to the surface via a swipe in

the opposite direction from contaminated gauze. The surface was then dry abraded with 400-grit silicon carbide paper in an effort to remove unwanted debris (in the case of the beam, it would be the oxidized metal). This process was then followed by a wet abrasion of the surface with 400-grit silicon carbide paper. The surface is first doused with a conditioner (M-Prep Conditioner A, Micro-Measurements, Raleigh, NC) and then thoroughly abraded while continually re-wetting the surface. The wet abrasion of the beam surface is a much more thorough and laborious process yielding a final surface that is polished and totally free of oxidation. The surface was then degreased once again in preparation for the application of the gauge layout lines. These lines served as a method of locating and orienting the gauge and were burnished onto the surface via a ballpoint pen or a brass tipped rod. The lines were burnished perpendicular to one another, with one line oriented in the direction of strain measurement. The gauge is then installed so that the triangular index marks defining the longitudinal and transverse axes of the grid are aligned with the reference lines on the test surface. After the gauge layout lines had been burnished, the surface was scrubbed with cotton tipped applicators and the surface conditioner (M-Prep Conditioner A, Micro-Measurements, Raleigh, NC) followed by thorough degreasing.

With the layout lines in place and the surface conditioned it was necessary to return the surface to an optimum pH of approximately 7.0 to 7.5. This was accomplished by wetting the surface thoroughly with a surface neutralizer (M-Prep Neutralizer 5A, Micro-Measurements, Raleigh, NC) and scrubbing with cotton tipped applicators. The beam surface was continuously wetted throughout this step since the neutralizing agent would deposit a residual precipitate onto the surface if allowed to dry. Finally, a single swipe of cotton gauze removed the excess surface neutralizer.

With the surface prepared, three 350 Ω stacked rosette gauges (WK-XX-060WR-350, Micro-Measurements, Raleigh, NC) were bonded to the surface. First, using degreased and neutralized tweezers; a single gauge was removed from the package and placed bondable side down on a chemically clean and neutral glass slide. Then a 4 inch piece of cellophane tape (PCT-2A, Micro-Measurements, Raleigh, NC) with non-removable mastic was placed over the glass slide taking care to ensure that the gauge was centered within the tape. Using a shallow angle and in a controlled manner the tape was lifted up carrying the gauge along with it. Figure 3.1 (adapted from http://www.vishay.com/brands/measurements_group/guide/ib/b129/129index.htm) illustrates this process.

Figure 3.1: Removal of Tape with Gauge Assembly Attached.

With the specimen firmly clamped to avoid misalignment, the tape was placed on the surface lining up the burnished layout lines with the triangular alignment marks present on the gauge itself. The tape was then lifted, exposing the gauge's bond surface, folded back over itself and neatly tucked forming a loop. In order to ensure even polymerization of the bonding, and expedite the curing time, a bonding catalyst (M-Bond 200 Catalyst, Micro-Measurements, Raleigh, NC) was applied to the gauge's surface. The procedure for applying the catalyst is shown in Figure 3.2 (adapted from http://www.vishay.com/brands/measurements_group/guide/ib/b129/129index.htm) and was similar to the degreasing procedure in that the brush was swiped in one direction only to ensure an even application of the catalyst. A minimum of one minute drying time was put in place after the catalyst was applied.

Figure 3.2: Application of the Adhesive Catalyst to the Gauge Surface.

After the catalyst had thoroughly dried, two drops of the methyl-2-cyanoacrylate adhesive (M-Bond 200, Micro-Measurements, Raleigh, NC) was applied to the bottom edge of the tape. The tape was then folded back over the specimen and a single swipe of cotton gauze across the back of the tape/gauge assembly ensured the even application of the adhesive as shown in Figure 3.3 (adapted from http://www.vishay.com/brands/measurements_group/guide/ib/b129/129index.htm).

Figure 3.3: Application of Adhesive to Specimen Surface.

This was followed by immediately pressing the back of the gauge firmly with finger pressure. The heat from the finger aids in the polymerization process. This pressure was maintained for a full five minutes before it was released and this was followed by another five-minute incubation period before the cellophane tape was removed for soldering.

For wiring the gauge assembly, single stranded 30 AWG wire wrap was used in conjunction with a chisel tipped soldering iron and a tin-lead alloy solder. Precise melting temperatures were achievable by this solder due to its eutectic alloy nature that discouraged the creation of brittle solder joints. After soldering, the leadwires and solder joints were protected by a layer of polyurethane (M-Coat A, Micro-Measurements, Raleigh, NC) and the entire assembly was allowed to dry for 72 hours.

Since strain gauge output signals are generally quite small, signal amplification and bridge completion was accomplished with a strain gauge amplifier (DS 2011, RDP Electronics, Wolverhampton, United Kingdom). Since each rosette is, in reality, a collection of three gauges each requiring its own means of amplification for strain measurement, a total of nine working amplification channels were required. Previous work with other types of transducers required slight modifications in the internal circuitry of some of the channels. In total, thirteen channels of bridge completing circuitry were included in the amplifier allowing for four redundant channels to serve as replacements should failure occur. Each channel was wired for $\frac{1}{4}$ Wheatstone Bridge completion and

both an oscilloscope and voltmeter were used as monitors for the output balancing of the gauges at an excitation voltage of 5V.

With the gauges in place and the electronics wired, the steel beam was then clamped firmly to a rigidly supported lab bench. Known weights were then placed on the end of the beam to induce transverse deflections and thus strains in the beam/gauges. Qualitative observations of the change in signal magnitude in response to the changes in known weights were then made and the loads removed.

3.1.2 Application of a Strain Gauge Assembly to Embalmed Bovine Ribs

The aim of this secondary exercise was to confirm the effectiveness of the strain gauge application procedure to a biological structure with significant curvature. Minor modifications to the procedure outlined in Section 3.1.1 were required to ensure proper bonding to a bony surface.

Two segments of adult bovine ribs were removed of all soft tissue and stored in a 10% formaldehyde solution. The ribs were retrieved from the formaldehyde solution and submerged in water the day of the exercise. Upon their removal from the water, the ribs were towed dry to remove excess water and then thoroughly degreased with isopropyl rubbing alcohol. The bone surface was both wet and dry abraded according to the guidelines in the Oxidized Steel Beam section though not nearly as aggressively since the unwanted debris on the bone surface was removed much more easily than the beam's oxidation.

Since the outer layer of cortical bone is often very thin and commonly contains many periosteal blood vessels, the area of application was sealed and prepared for the adhesive (Szivek & Gharpuray, 2000). After the surface was conditioned and neutralized, a layer of adhesive catalyst was applied to the surface and allowed to dry for five minutes. After drying, a few drops of methyl-2-cyanoacrylate were applied to the surface and spread uniformly with a cotton tipped applicator and allowed to dry for five minutes. This was followed by a period of light dry abrasion with 400-grit silicon carbide paper to smooth out any imperfections in the adhesive's outer surface as a result of uneven drying. The newly formed surface was then conditioned and neutralized and the gauges put in place according to the aforementioned procedure outlined in the Section 3.1.1. The only other modification to the procedure involved the used of thumb pressure instead of finger pressure as shown in Figure 3.4. The larger surface area provided by the thumb ensured that the corners and edges of the gauge assembly conformed accurately to the ribs curved surface without any gaps or creases.

Figure 3.4: Application of Strain Gauge Assembly to Embalmed Bovine Rib.

After a five-minute drying period with thumb pressure in place to aid in the polymerization, the ribs were wrapped with rags soaked in water and placed into a sealed plastic bag for transport. Twenty-four hours later, the ribs were removed from the plastic and thoroughly examined for bonding. The final strain gauge assembly as bonded to the embalmed bovine rib is shown in Figure 3.5.

Figure 3.5: Final Positioning of Strain Gauge Assembly on Embalmed Bovine Rib.

The examination process consisted of a magnified examination of the gauge-bone interface in an attempt to reveal gaps and or folds. A gauge that has poorly conformed to a curved surface will exhibit gaps, folds and possible air bubbles. The mechanical three-point bending of the rib with fulcrum under the site of attachment (on the opposite side) was also performed in an attempt to observe any peeling of the gauge from the bone's surface.

3.1.3 Mock Experiment Using a Plastic Model Skull

The aim of this preliminary test was to establish a standardized procedure for the actual impact experiment to be performed on cadaver maxillæ, as well as evaluate the data acquisition and motion capture systems.

3.1.3.1 Design and Creation of the Data Acquisition System

In order to properly capture the mechanical response of the impact, a sophisticated data acquisition system was developed. This system required the appropriate voltage range for the transducers involved, a sampling rate fast enough rate to capture the peak load of the impact and be time coded so that a distinct beginning and ending to data acquisition occurred. While commercial software is available that may expedite the data acquisition process, steep learning curves are associated with the customization of such programs and thus it was decided that a system built from the ground up was more applicable to this study.

Firstly, the output signal from any transducer, generally speaking, is analog in nature. By default, a computer (and consequently a computer aided data acquisition system) is invariably digital. Therefore, a translation from analog to digital is mandatory. The rate at which the analog to digital converter operates is clearly the limiting factor in any data acquisition system. Analog data coming from the transducer amplifiers (load cell and strain gauge) was first sent to a break-out box that relayed the signals through a pair of 16 pin ribbon cables and then converted to digital data via two 16-bit A/D data acquisition cards (PC-CARD-DAS16/16AO, Measurement Computing Inc, Middleboro, MA). Each card housed sixteen separate channels with a single channel maximum operating rate of 100kHz.

With the hardware in place, custom written software was designed using MATLAB V6.5 (Mathworks Inc, Natick, MA). The primary task of the data acquisition software was to create a series of data acquisition channels, each capable of receiving an incoming analog signal. This incoming signal was translated to a digital format and stored safely in an array that was sized

according to the number samples taken. The name of the array was unique for each session of acquired data. Since more than one type of transducer was employed in the experiment, each transducer's data was stored in a separate file.

As a secondary function, the data acquisition system provided a means of monitoring a successful data acquisition session. The simplest way of accomplishing this was to plot the incoming voltage versus the number of samples taken as soon as the session was completed. This was the technique employed in the impact experiments.

The design of the software also included specification of the voltage range for each transducer, which was $\pm 10\text{V}$. This specification is important because should the voltage range be exceeded, an abrupt clipping of the signal would occur at values above the upper threshold causing a loss of all data above that value. On the other hand, setting the voltage range at too high a range would induce a loss of accuracy within the signal itself. This is analogous to trying to pick up a penny with a thickly padded leather glove; it is simply too large to be able to negotiate the subtle difference between the penny and the surface for which the penny rests on.

Following the specification of the voltage range, an appropriate sampling rate was chosen. The selection of sampling rate depends on the mechanical response of the system and the maximum rate for which the acquisition system will allow for multi-channel sampling. The mechanical response of dry human skulls is approximately 587.5 Hz (Yasuyoshi *et al.*, 1992). According to the Nyquist criterion, in order to successfully capture an event by sampling without aliasing, a sampling frequency (i.e. rate of data points captured) must be two times larger than the maximum frequency to be measured. If a sampling rate is not chosen to be at least twice the maximum value, an erroneous capturing of the event will be the result. In the case of the impact experiment, the resonating frequency of the head is the maximum frequency to be measured and therefore, by virtue of the Nyquist criterion, the sampling rate would have had a value of 1175 Hz. However, the maximum multi-channel sampling rate of the available data acquisition cards was 6750 Hz, and therefore this value was chosen since a higher sampling rate (in this case the highest available) would invariably capture a larger number of data points and yields a much smoother response profile for the impact. The duration of a high velocity-low mass impact to the maxilla is generally quite small, approximately 0.44 milliseconds or longer (Crisco *et al.*, 1997), and therefore a sampling rate of 6750 Hz captured with a minimum of three data points, the precise moment of impact as well as the resonance afterwards.

The duration of data acquisition was selected next and this was based on the approximate duration of the entire experiment. Keeping in mind the required time for the impactor to be accelerated to the desired speed, as well as traverse the distance from its release point to the point of contact, a value of 5 seconds was chosen. The triggering of data acquisition allowed for pre and post sampling of the signal and this technique was employed to ensure that if any errors in timing occurred there would not be an accompanying loss of data. It was important that post impact time be allocated should there be any vibrational responses of the system which required time to dissipate.

As a matter of operator convenience, all of the above functions were automated within the data acquisition software, with the exception of the actual naming of the log file to which the data array was stored. This was accomplished by the operator just prior to the impact experiment and triggered by the command prompt entry “ready(‘filename1’, ‘filename2’)” and the actual data acquisition session was triggered via the command prompt entry “fire”.

3.1.3.2 Design and Implementation of Motion Capture System

In order to fully examine the nature of the deformations involved in the impact experiment, a pair of high-speed 16mm cameras (Nova 16-3, Photo Kinetics Inc, New York, NY) captured the entire experiment from both frontal and lateral views (see Figure 3.6). In order to establish a time reference for the film frames, the high-speed cameras were triggered through the movement of the impactor by a photosensitive diode, which also triggered a 1 kHz sync pulse that imprinted itself onto the actual film. This gave a definite beginning to the experiment and a precise cadence for its duration.

Figure 3.6: Nova 16-3, 16mm High-Speed Camera.

The appropriate lighting required to capture the impact experiment at high speed required the selection and calculation of several parameters. The most important of the calculated parameters was the frame rate, is a time-based quantity that describes the number of frames passing through the camera's aperture in a given unit of time. In general the units for frame rate are given as [frames/second] and this was the convention followed. To calculate the required frame rate, the following equation must be used:

$$F.R = \frac{C \cdot S \cdot \cos \theta}{W \cdot B} \quad (3)$$

where C is a constant for 16mm cameras having a value of 200. This is based on an image blur of 0.002 inches present at high speeds but does not show at normal speed or frame-by-frame viewings of the film. S denotes the subject velocity in units of [inches/second] and was based on the impactor's speed. Since the velocity of the impactor for the duration of all impact experiments was intended to be 14m/s this translated to a value of 880 in/s. The unknown θ represents the angle between subject motion and the axis of the camera's lens and since the camera was aligned along the direction of the impactor's trajectory it was assigned a value of 0 radians. W denotes the field width, which was arbitrarily chosen to be 12 inches. This value was based on the dimensions of the cadaver head, the head fixture and for the allowance of a few inches of impactor travel in order to confirm the impactor's incoming velocity through a time-coded frame-by-frame analysis. Finally B denotes the camera's shutter ratio, which was assigned a value of 10:1. This ratio means that 10% of a full 360-degree view will be used in the image, which translates to, 36° or 0.63 radians. Again this was chosen based on the angular dimension of the cadaver head, the fixture and some travel distance as measured by a camera-to-subject distance of approximately two metres. Taking all the aforementioned values into consideration, a value of 1,466.67 frames/second was calculated as minimal frame rate. Keeping in mind that a 110 ft reel of film was to be loaded into the camera, and that the motors driving the reels are constantly accelerating, the actual film rate achieved for the motion capture was approximately 3,200 frames/second at the end of the reel and approximately 2,400 frames/second at the moment of impact.

With knowledge of the approximate frame rate for the experiment intact, the actual illumination required for the motion capture could now be calculated via the equation:

$$I = \frac{K \cdot A^2}{ISO \cdot ET} \quad (4)$$

where K is a constant with value 25, A is the aperture setting for the camera with an assigned value of $f/4.0$, ISO is the corresponding film speed of 500 and finally, ET is the exposure time of the film in seconds. The value for ET was set at 4.167×10^{-5} s or $1/24000^{\text{th}}$ of a second. With all of the parameters set, the minimum value of illumination was calculated to be 19,203.07-foot candles or $20.67 \text{ Lumens/cm}^2$. That translated to no less than four 1000 Watt PAR-64 sealed beam spot lamps 120 VAC power at a distance of two metres from the object to be illuminated.

3.1.3.3 Design and Manufacture of Skull Fixation Device

In order to ensure that the kinetic energy of the impact be solely absorbed by the body to be impacted, a rigid fixation device designed to hold the head/skull firmly in place without sliding and without obstructing the maxilla was required. Such a device did not exist and therefore a custom design was required.

First, an aluminum Halo frame with numerous mounting holes was obtained and used as the basis for the design. It's semi-elliptical shape allowed for the mounting of heads of various sizes. The remainder of the fixture consisted of a series of perpendicularly oriented, mild steel plates and beams that served to support the halo at its temporal and occipital mounting holes (see Figures 3.7, 3.8 and 3.9). The lower portion of the device housed a six-degree of freedom load-cell (PY6-300, Bertec Corporation, Columbus, OH) intended to measure three force (F_x , F_y , F_z) and three moment (M_x , M_y , M_z) components for the entire structure. The output from the load cell was then amplified and conditioned (AM6110, Bertec Corporation, Columbus, Ohio) and passed through a second-order 1 kHz filter to remove any unwanted noise from the signal. Mounted beneath the load-cell, a steel plate lined with a layer of synthetic rubber to increase friction was clamped down to a rigidly supported lab bench.

An aluminum tube, bent and flattened into a U shape was temporarily mounted to the front of the head fixture with removable screws. This tube would hold both the acrylic panel and foam backing used in the accuracy control process. Once the accuracy was confirmed, the tube, panel and foam backing were removed.

The entire head fixture was then spray painted with a black, matte finish in order to minimize reflection from the lighting equipment. On one side of the fixture, a cork bulletin board was clamped in order to provide a backdrop for the lighting and filming. The board was first spray painted black with the same matte finish, and then a series of reflective dots placed 2.54 centimetres (1 inch) apart

Figure 3.7: Top View of Head Fixation Device.

Figure 3.8: Side View of Head Fixation Device.

Figure 3.9: Head Fixation Device with Cadaver Specimen Mounted.

in a Cartesian grid fashion. These dots represented landmarks of known locations and simply served as a ruler for the high-speed filming.

The model head itself was fixed into place with ten custom-made 3" stainless steel screws with an ultra fine thread. The fine thread ensured minimal tissue (or plastic) disruption occurred during the mounting of the head. The manufacturing of all components and all tig welding of joints was completed on site at the Bloorview Macmillan Children's Centre, Toronto, ON.

3.1.3.4 Impactor Selection and Velocity/Accuracy Control

In order to replicate the kinds of blunt force trauma treated by clinicians, the judicious selection of an impactor was crucial. It was decided that a major league quality baseball (Rawlings Official Major League Baseball, Rawlings Sporting Goods, Fenton, MO) with a mass of 150 grams was an ideal impactor because both its accuracy and velocity could be well controlled.

The testing protocol utilized a commercial ball-pitching machine (Ace '99 Pitching machine, American Athletic Inc., Jefferson, IA) and is shown in Figure 3.10. The output velocity of the machine was rated to be as high as 90 mph (145 km/h), which was much higher than required to induce the kinds of blunt force trauma seen by clinicians. Therefore a speed of 50 mph (80 km/h) was arbitrarily decided on for the impact to the plastic skull. The output velocity from the pitching

Figure 3.10: Ace '99 Commercial Pitching Machine.

machine was controlled via a set of two dials governing the rotational velocity of the two wheels. This coarse speed adjustment was verified using a radio frequency Doppler monitor (Speedchek, Outer Limits Inc., Edina, MI).

As a consequence of the design of the pitching machine, the output velocity was a function of the input velocity and therefore a constant input speed must be maintained if a constant output speed is desired. The constant input velocity was achieved by modifying the pitching machine slightly. A series of cast iron rings preceded the two counter spinning wheels, which served as the propulsion mechanism for the baseball. These rings accounted for the use of softballs as well as hardball baseballs. Since our experiment did not require the use of softballs, the outermost ring was removed from the machine and a 4-inch flexible tube was attached to the inner ring that was directly adjacent to the counter spinning wheels. When raised to a known height and kept there, the baseball could be dropped through the flexible tube and fed into the counter spinning wheels at a consistent speed due to the acceleration of gravity.

With the velocity control established, mounting a simple laser pointer to the pitching machine controlled the accuracy of the impact. The location of the pointer was calibrated by firing a series of balls at a foam mat dusted with talcum powder and shimming the laser pointer until the beam focused precisely on the centre of the circular dust mark created by the impacting ball. As a secondary means of accuracy control, the ball machine was placed two meters away from the skull fixation device, with a plastic model skull fixed firmly in position. Placing a sheet of acrylic, backed with foam and dusted with talcum powder, the ball was repeatedly fired until the accuracy and velocity were correct. The acrylic sheet was then removed and the pitching machine was ready to pitch the ball.

3.1.3.5 Preparation of the Plastic Skull

A plastic model of the human skull (Budget Skull, Dick Blick Art Materials, Galesburg, IL) was purchased and the mandible was removed. The normally detachable cranium was glued securely into place with a twenty-four hour epoxy. After drying the excess residue from the gluing process was abraded with 400-grit silicon carbide paper until smooth. It was decided that a single 350Ω stacked rosette strain gauge (WK-XX-060WR-350, Micro-Measurements, Raleigh, NC) placed on the *medial* palate of the skull would be suffice to capture the profile of the impact and show the effectiveness of the data acquisition system. With the gauge bonded in place according to the same protocol as employed in the bovine rib exercise, six single stranded 30-AWG wire wrap leads were soldered to the individual terminals of the gauge and relieved for strain via a taped loop. The plastic skull was then placed into the head fixture and fixed into position with the ten custom made stainless steel screws until it was rigidly supported. With the skull in place, the output cable from the load cell was attached to the load cell amplifier/conditioner. Three, 24-AWG, individually shielded wires were soldered to the lead wire pairs of the gauges, looped for strain relief and wired into the strain gauge amplifier exciting the gauges to 5V. Each direction on the rosette received its own independent amplification channel. The entire structure was then placed onto a lab bench and firmly clamped into position.

3.1.3.6 Experimental Procedure

Due to the large amount of current drawn by the pitching machine, the high-speed cameras, and the lighting equipment, the power supplied from wall outlets for the experiment was divided into several groups. Each group was supplied with power from an outlet belonging to a different circuit in order to reduce the possibility of a power failure occurring because of high current draw.

The strain gauge amplifier and the load cell amplifier were both turned on and allowed to warm up for an hour to ensure steady state was reached. All computer systems and data acquisition hardware were also powered and booted up.

The baseball machine was placed directly in front of the lab bench holding the head fixture at a distance of 2 metres. At the rear of the machine, a flexible tube with a diameter of 0.1016 metres (4 inches) was attached and its opposite end was attached to a wooden stand 170 centimetres (67 inches) tall. A rectangular opening was cut into either side of the raised end of the tube in order to place the photosensitive diode that would trigger the motion capture system. Breaking the light beam's path with a moving hand tested this diode. A baseball was dropped through the tube to verify that the path was free of obstructions and that the ball was able to reach the counter spinning wheels without any outside assistance.

The motion capture system was placed perpendicularly to the trajectory of the baseball at a distance of 2 metres in order to provide a lateral view of the impact. To reduce costs generated by purchasing and developing film, only a single high-speed camera was used for the mock experiment. The camera was then focused onto the field of view using a printed calibration table (Kodak Canada Inc., Toronto, ON), which consisted of a series of parallel and perpendicular lines. When all lines appeared sharp, the resulting image is deemed focused. A single 33.5 metre (110 foot) reel of black and white film (ISO-500, Kodak Canada Inc., Toronto, ON) was loaded into the camera checked for tension by small rotation of the reel.

With the camera in place, the lighting equipment was then placed around the camera at a distance of two metres, and the beams of light were focused directly onto the head fixture. The intense heat generated by the lights would cause an increase in temperature of the specimen and lead to inaccurate strain gauge values and therefore were left off until just before impact (Szivek & Gharpuray, 2000).

The output cables from both the strain gauge and load cell amplifiers were then attached to the custom-built break box. From there, the two, 16-pin ribbon cables were attached to the data acquisition cards and the cards were installed into the computer. The software was then activated, which confirmed that incoming data could now be acquired.

The pitching machine was then turned on and the speed settings were set at 5 for the upper wheel and 7 for the lower one. While the wheels were allowed to reach their maximum rotation speed, the aluminum tube and acrylic panel with foam backing were then put into place on the head fixture. Preliminary height adjustments to the pitching machine were made using the laser pointer as a guide.

Dusting the acrylic panel with talcum powder, and placing the Doppler monitor next to the head fixture, the speed and accuracy of the baseball were repeatedly checked and adjusted until a consistent velocity and impact location was achieved. The aluminum tube, along with the acrylic panel and foam backing were then removed from the head fixture.

The strain gauges were then balanced with both an oscilloscope and a voltmeter, and the load cell was zeroed. Signal strength and response was quickly checked by pressing on the maxillary incisors of the plastic skull and observing changes in the oscilloscope and voltmeter readings. Having checked the signal, the gauges and load cell were then rebalanced and rezeroed.

The names of the log files for which data would be recorded were then specified and the data acquisition software was initiated and cued for manual triggering. The lights were then powered, the software triggered and the ball was dropped through the tube activating the motion capture system and impacting the model skull. The data was recorded and saved to file and a series of plots showed the impact profile. After the impact, the skull was removed from the fixture and examined for a qualitative description of the fracture(s).

3.2 Impact Experiment with Cadaver Heads

The final impact experiment involved the use of six (5 male and 1 female) embalmed human cadaver heads. The mandibles were not required in this study and thus were removed from the heads. The specimens were not necessarily fully edentulous and only required that the four *central* maxillary incisors and 2 canines be intact. Human anatomical material was supplied through the Division of Anatomy, Department of Surgery, at the University of Toronto under ethical approval of the Chief Coroner of Ontario and the Research Ethics Boards of both Bloorview MacMillan Children's Centre and The Hospital for Sick Children. All experiments were performed at the laboratories of the Division of Anatomy and CT scans were performed at Princess Margaret Hospital. Material was transferred between sites by a funeral director licensed in the province of Ontario.

3.2.1 Preparation Of Cadaver Heads

The cadaver heads were first placed onto the open end of a polyvinyl (PVC) tube as a temporary mount, which prevented any movement of the head and allowed for the positioning of the head to occur hands free. Rotating the head so that the *inferior* surface of the palate was facing upwards the removal of the soft tissue began just behind the *palatal* surface of the *central* incisors and went as far *posteriorly* as the *palatal* surface of the third molars. Using a scalpel and periosteal

elevator, soft tissue was removed taking care not to damage the underlying bone. With the tissue removed, the bone surface was cleaned and disinfected with isopropyl rubbing alcohol.

Rotating the head so that the maxillary incisors were facing upward, it was decided that two of the specimens would have all of their soft tissue surrounding the maxilla removed. This included the *labia*, periodontal tissues and the lower portions of the nose. The remaining four heads were to have their periodontal tissues excised only. All other soft tissues were reflected upward to apply instrumentation and returned to their natural position as closely possible. Again, using a scalpel and periosteal elevator, the soft tissue was removed and the bone surface underneath was cleaned and disinfected.

The locations of the strain gauges were based on physiological landmarks present on the maxilla and palate. Two gauges were to be placed *bilaterally* on the *canine fossae* and one to be placed just *posterior* to the *anterior palatine fossa* on the *medial* palate. This ensured that although different specimens were to be tested, the relative locations of strain measurement were approximately equal.

The same procedures and apparatus used in the mounting of strain gauge assemblies to bovine ribs were applied to the human maxillae with a few modifications. The specimens were thoroughly degreased and abraded, although more care was taken since the *anterior* surface of the maxilla just above the *canine fossa* tends to be quite thin and can be punctured during abrasion. To ensure good adhesion to the *inferior* surface of the palate, which is littered with *foramina*, a bed of catalyst and adhesive was created in similar fashion to the bovine exercise only thicker. As a final modification, each gauge was to be covered by a flap of dental masking bonded just above the gauge to prevent any fluid from corrupting the gauge.

With the gauges successfully in place, six, 30-AWG, single stranded wire wrap leads were soldered to each of the three rosettes. These leads were taped off in pairs corresponding to each of the three gauges present on each rosette for a total of 18 leads. The taped pairs were labeled left (L), centre (C) or right (R) corresponding to their positions within the rosette relative to the head facing an observer. The locations of the gauges themselves were numbered from one to three. Beginning with the gauge above the subject's right *canine fossa* as one the gauge on the *medial* palate was two and finally the gauge above the patient's *left canine fossa* was three.

The heads were wrapped in rags soaked in an embalming solution containing isopropyl alcohol, propylene glycol, amphyl, liquefied phenol and 10 per cent buffered formalin. The heads

were then placed into plastic containers and allowed to rest for a minimum of twenty-four hours in a chilled morgue.

3.2.2 Experimental Procedure

The same procedures and apparatus used in the mock experiment with the plastic model skull were also applied in the cadaver impact experiments, with a few modifications. On the day of testing, the heads to be impacted were removed from the chilled morgue and taken out of their plastic containers allowing them to reach room temperature. The preservation of the soaked rags on the heads maintained their moisture until they were to be tested. With all the experimental equipment in place according to the mock experiment protocol, the head to be impacted was removed from the rags and the resistance of all gauges was tested using an ohmmeter. The values given by the ohmmeter were compared to their manufacturers' specifications in order to evaluate whether or not any residual strains had been induced in the gauge during application. Using a multi-meter, the ends of the leads were then checked for current flow indicating an unobstructed flow of current through the leads, the gauges and the soldering points. Since the experiment was going to be filmed and photographed for documentary purposes, the identity of the cadavers had to be preserved. This was accomplished by covering all portions of the head, except for the most *inferior* portions of the nose and the maxilla, with a hair garment commonly referred to as a "du-rag" (Du-rag tie down cap, Bee Sales Co., Chicago, IL).

The head was then placed into the fixture and secured firmly with the stainless steel screws. The strain gauge leads were looped for strain relief and taped to the head fixture, which was then clamped securely to a lab bench. The precise locations of the strain gauges were manually digitized using a Microscribe 3D digitizer (Microscribe 3D, Immersion Corporation, San Jose, CA) as shown in Figure 3.11 (Adapted from <http://www.immersion.com/digitizer/products/index.php>). Points were digitized using the lower left corner of each gauge as a landmark and a rigid point on the fixture for the co-ordinate system's origin. The lead wires and output from the load cell were wired into their corresponding amplifiers using the protocol from the mock experiment. The gauges and load cell were then balanced and tested for signal strength and quality.

The pitching machine and the flexible tube were placed two metres away from clamped head fixture facing head on. Two identical high-speed cameras were used this time with the first camera and the lights *laterally* placed approximately one metre away from the target. The second camera was

placed next to the pitching machine and was slightly off the perpendicular to the head fixture. Figure 3.12 illustrates the positions of some of the equipment used and the trajectory of the baseball.

The velocity and accuracy of the impact were controlled via the mock experiment protocol. The only modification to this procedure was a lowering of the impactor's velocity. Instead of using 22.35 metres/second, the value was lowered to 14 metres per second. The decision to lower the impact velocity was based on the fact that professional athletes can throw a baseball over 40 metres/second, youths at approximately 26 metres/second and little league coaches set their pitching machines at 14 metres per second (Fleisig *et al.*, 1999). The lowest setting was chosen in order to keep the trauma localized to only the maxilla and the maxillary incisors since fractures to the maxillofacial complex often occur in combination with other facial fractures (Richardson, 2000). To achieve this speed the dials on the pitching machine were set to 4 for the upper wheel and 6 for the lower. These values were later adjusted to 3 and 5 respectively during the acrylic panel accuracy testing.

Figure 3.11: Microscribe 3D Manual Digitizer.

Figure 3.12: Lateral View of Cadaver Impact Experimental Setup.

With the equipment in place and the software cued, the gauges and load cell were rebalanced and rezeroed and the temperature of the room and cadaver were measured using a laser temperature probe (Actron III Infrared Thermometer Non-Contact CP7872, Actron Manufacturing Co., Cleveland OH)). The software was triggered and the ball was dropped activating the motion capture system and impacting the cadaver head. The data was then logged to file and saved and the specimen was removed from the head fixture to be examined and photographed. The examination involved the charting of all injuries, fractures, avulsions etc. according to the FDI convention. All anatomical matter that was completely detached from the head (with the exception of any avulsed teeth) was gathered and placed with the head and returned to the plastic container, again wrapped in rags soaked with embalming solution. The entire experimental procedure was then repeated for the remaining cadaver heads.

Chapter 4

4 MODELING METHODOLOGY

4.1 Creation of the Solid Model

The objective of this portion of the study was to create a geometrically accurate model of the human maxilla and the maxillary incisors. In order to accurately capture the complex geometries present in a human maxilla, a non-destructive imaging technique was used. The most logical choice for this imaging technique is computerized tomography (CT), which enables the researcher to accurately depict the tissues present in the maxilla. From the captured images, a solid, three-dimensional model of the maxilla could then be reconstructed and rendered.

4.1.1 Computerized Tomography (CT)

CT scanning is a non-destructive imaging technique created through the use of x-rays and computers. The image is produced by bombarding the object in question with a series of narrowly focused beams of x-rays. Opposing these beams are numerous x-ray detectors that measure the intensity of each incoming beam after it has radiated the object. The strength of an incoming beam depends on the x-ray attenuation of the material obstructing its path. As the x-ray/x-ray detector assembly is rotated around the object, numerous beams are radiated, attenuated and detected which the computer then processes. The processing technique mathematically creates an image in which the resulting pixel intensity of a point is directly proportional to the radio-intensity of a beam at that precise location within the object (Freedman, 1988). Since the beams radiate through the object transversely, each single CT image is a “slice” of known thickness through the object at a specific height. By repeating the technique at numerous heights, evenly spaced apart, a complete scan of the

object is possible. Assembling these slices into a stack (analogous to a sandwich being made) will reveal the complete geometry (internal and external) of the imaged object.

On the day of imaging, a fully endentate pediatric skull (female, age 12 years) as shown in Figure 4.1 was placed on the table of the CT scanner (General Electric LightSpeed RT Fly-Set Scanner, General Electric, New York, NY) such that the plane of section passed perpendicularly to the plane created by the most inferior portion of the mandible. Slices were placed 0.62mm apart, yielding a total of 461 images, which was the highest resolution setting for the scanner. This level of accuracy was required in order to capture fine geometric detail that would be missed with a coarser setting. Each image was produced at 1.5x magnification and included within the images was a calibration bar that provided a relative measure for the magnitudes present within the image.

Figure 4.1: Dry Skull Specimen Used in CT Imaging.

4.1.2 Image Processing, Boundary Detection and Segmentation Techniques

The output from a CT scanner is an array of files numbered sequentially as they are created. The storage format of choice is the Digital Imaging and Communications in Medicine (DICOM) format created by the National Electrical Manufacturer's Association (NEMA). A single DICOM file contains both header information, which is patient and scan specific information regarding dimensions and image type, and the actual image file itself. In order to view the outputted DICOM files, specific software was utilized (3D Doctor, Able Software Corp., Lexington, MA) to manipulate the images through a user interface. An example of a raw DICOM file is shown in Figure 4.2.

Figure 4.2: Raw CT Image in DICOM Format.

The processing of the images was organized with a top-down hierarchy of components and tasks. The first task was the organization of the images at the project level. This involved loading the image stack into the viewing software and the examination each image closely. If an image was of poor quality, or at a location within the object that we were not interested in, the image was removed from the current project. For example, since the scope of this study is focused on the maxillofacial complex, any images of the mandible were not required and therefore removed.

Images that were of poor quality were deconvoluted using a maximum entropy based algorithm in order to restore their quality. The performance of this deconvolution algorithm was so impressive that all the images were then subsequently passed to remove any “fuzziness” caused by the imaging process.

The second task of the image processing protocol was the detection of the boundaries present and their subsequent segmentation within each image. Since the image is a transverse slice of the anatomy, boundaries between individual bones and cavities would be present and required identification. The segmentation procedure employed was an interactive, threshold-based algorithm. First, an image slice containing numerous boundaries and bodies was selected. Then, a region of interest (ROI) within this slice that equally exploited both dark and light pixels was selected. By defining a threshold for the grayscale-value of the pixels within the ROI, the algorithm was trained to remove any pixels that were below this assigned value. With this technique, individual, closed boundaries of the outer structures are distinguished, as well as the boundaries of cavities within the

bone structure present in the ROI. Having trained the algorithm, the ROI was expanded to cover the entire image and the segmenting procedure repeated. Unfortunately, it is nearly impossible to discover a ROI that will accurately represent the distribution of pixel gray values for the entire image, and therefore a manual editing of the created boundaries ultimately followed this segmentation procedure.

The lines created in the segmentation are, in reality, polylines interpolated by third order splines through nodal points. These splines can be manipulated through the relocation, redistribution and renumbering of these nodal points. The ideal situation is one in which the geometry of the boundary is accurately met with the fewest number of influential nodal points. A continuous, smooth curve should be the result of such a situation and that was the goal of the manual editing protocol. First, any extraneous nodes were removed by tracking their influence on the curve. If a node did not influence the curve largely, relative to its neighbouring nodes in a qualitative sense, it was removed. This process, known as node reduction, drastically reduces the overall complexity of the geometry. Once the nodes had been reduced, the spline was recalculated and redrawn through the nodes yielding a new boundary. If the boundary still contained gross discontinuities due to flying pixels, a new ROI was established and the threshold value was lowered until the extraneous pixel was no longer influential. If the previous technique was ineffective, the threshold was returned to its original value and the node associated with the flying pixel was relocated closer to the actual boundary and the spline was redrawn. Finally, regions of high curvature require more nodes than linear regions in order to capture the geometry through interpolation. Thus, without altering the overall number of nodes, a redistribution of their location within the boundary may capture the topology more accurately at no mathematical expense (Able Software Corp, 2003).

The outlined interactive segmentation procedure and the subsequent manual editing of the boundaries were then repeated for all image slices present in the project stack. With all slices interactively segmented, an inter-slice connectivity algorithm was employed to verify that the geometry of adjacent slices was compatible and free of discontinuities. This ensured that there were no sudden changes in the boundary data and that all boundaries were interconnected throughout the slices verifying the structure's closure. Upon the completion of the interconnectivity algorithm, the project stack was saved and then examined by dentists' (Dr. David Kenny, D.D.S, PhD and Dr. Michael Casas, D.D.S, MSc) to verify the geometric representation of the chosen anatomy. A sample of a typical CT image in DICOM format after its boundaries have been segmented is shown in Figure 4.3.

Figure 4.3: CT Image With Boundaries Segmented.

4.1.3 Manual Creation of Non-Uniform Rational B-Spline (NURBS) Surfaces from Exported Boundary Curves

In order to create a three-dimensional volume reconstruction from the boundary data, appropriate rendering software was required (Rhino V2.0, Robert McNeel and Associates, Seattle, WA), as well as a suitable file medium. Initially, all the available boundary contour data was reconstructed within 3D Doctor™ as a simple surface and exported as a stereolithography (.STL) file. This was then imported into Rhino™ and volume rendered. The resulting rendering was extremely complex and its file size was approximately 1.67 gigabytes, which was wholly impractical. In order to reduce the file size, the individual boundary curves were exported as Initial Graphics Exchange Specification (IGES) files (see Figure 4.3), which is a neutral file type importable by many Computer Aided Design (CAD) software packages. From here, the boundary curves were scaled down by 50 per cent to nullify the magnification created by the CT scanning process and the process of manually creating a NURBS surface network could then begin. The rationale behind modeling NURBS curves and surfaces will be briefly explained in the following paragraphs.

Figure 4.4: Imported Maxillary Boundary Curves.

A NURBS curve is a technique for modeling and approximating irregularly curved geometries. Mathematically it is described as a vector-valued piecewise rational polynomial defined as the ratio of two non-rational generalizations of Bézier basis functions known as B-splines (Piegl & Tiller, 1997). The shape of a NURBS curve is influenced by the location and orientation of its control points. The control of a NURBS curve is so effective that they can be used to represent conics sections exactly (Piegl & Tiller, 1997). The equation for a general NURBS curve is given is shown below:

$$C(t) = \frac{\sum_{i=0}^n N_{i,p}(t) w_i P_i}{\sum_{i=0}^n N_{i,p}(t) w_i} \quad (5)$$

where p is the order, $N_{i,p}$ is the B-spline basis function, P_i is the control points and

P_i^w represents the last ordinate of the homogenous point P_i with weight w_i (Piegl & Tiller, 1997).

Figure 4.5 shows a typical NURBS curve and the underlying polylines connecting its control points.

The creation of NURBS curves sets the basis for the modeling of surfaces lofted through these curves. Essentially, they are the wireframe for which the surface is “draped” over much like the curved rods in an open umbrella. Each individual curve influences the shape of the surface lofted through it, and will continue to influence its shape over an area outside its own dimension. The

degree of influence depends on the amount of curvature present in the curve itself. Should a sharp discontinuity in the curve occur, this will appear as a sharp spike in the surface and will decay over increasing distances as the abruptness of the discontinuity is increased.

Figure 4.5: A Typical NURBS Curve with Control Points Visible.

In general, a NURBS surface of degree (p,q) is given by the following formula:

$$S(p,q) = \frac{\sum_{i=0}^m \sum_{j=0}^n N_{i,p}(u) N_{j,q}(v) w_{i,j} P_{i,j}}{\sum_{i=0}^m \sum_{j=0}^n N_{i,p}(u) N_{j,q}(v) w_{i,j}} \quad (6)$$

where $N_{i,p}$ and $N_{j,q}$ are the generalized Bézier basis functions, $P_{i,j}$ is the control points and $P_{i,j}^w$ represents the last ordinate of the homogenous point $P_{i,j}$ with weight $w_{i,j}$ (Weisstein, 2004).

To build the geometry of the maxilla using NURBS surfaces, the boundary data from the CT scans (now converted to NURBS curves) was first discretized into a finite number of evenly spaced points along their curves. Regions of high curvature were assigned a larger number of points while regions of decreased curvature were not. The discretization of the boundary data allowed for the interconnection of the separate slices of NURBS curves along the z-axis of the boundary data “sandwich”. By interpolating through the discretized points, a curve network operating in two directions was then built as shown in Figure 4.6. Using the curve network as a skeleton, this allowed for the lofting of quadrilateral surface segments that could be merged together in a piecewise fashion

analogous to the creation of a patchwork quilt. Figure 4.7 shows some of these lofted surfaces as they are being mated together to form the final solid model.

Figure 4.6: NURBS Curve Network Created From CT Data.

Figure 4.7: NURBS Surface Network Lofted from CT/NURBS Contours.

It was necessary to segment the creation of the NURBS surfaces in order to preserve the integrity of the geometry. Larger surfaces covering hundreds of networked curves are limited in their

ability to negotiate all of the underlying B-spline basis functions and therefore will only loft through them in an average sense. The result is an overly smooth function whose boundaries are exact but inner domains favour no specific basis function(s). In an attempt to conform to the entire network of basis functions (along the directions *u* and *v*), the surface segments were limited to approximately 30 to 40 curves in the both the *u* and *v* directions of the surface. To ensure smoothness without the expense of excess computation, third order splines and surfaces were used for the entire curve/surface network.

Initially, the entire maxillofacial structure was reconstructed moving as *superior* as the cranium. However, as was the case previously, the inclusion of so much extra tissue led to the exponential growth of the associated file size and therefore it was decided that only the maxilla itself would be included in the final solid model. The modified surface network began at the most *superior* portion of the *superior* maxilla and worked as far *posterior* as the edge of the hard palate. A rendering of the final solid model is shown in Figure 4.8.

Figure 4.8: Final Rendering of Solid Maxilla Model.

The surface model was completely closed and was free of any naked edges making it a watertight boundary represented (b-rep) solid. In its final state, the solid model incorporated 4 separate maxillary incisors and 2 *buccal* tooth segments from canine to second molar. The geometry of the 4 maxillary incisors created by the CT images was somewhat blurry and inexact, and therefore

these teeth were manually reconstructed through the manipulation of the NURBS curves. The anatomical integrity of the modified geometry was confirmed and verified for completeness by an expert dentist (Dr. David Kenny, D.D.S, PhD).

In order for the model to be mathematically valid, the surfaces enclosing the solid must be free of kinks and all direction vectors must be pointing in the same direction. FE meshing and solving is not possible if the model contains such errors and therefore it is vital that it is completely void of these anomalies. After the solid model was complete, dental and engineering experts (Dr. David Kenny, D.D.S, PhD, Dr. Michael Casas, D.D.S, MSc., Dr. Kamran Behdinan, P.Eng, PhD. and Alan Morris, P.Eng, MAsC.) once again verified it for anatomical accuracy and completeness.

4.2 Creation of the Finite Element (FE) Model

FE analysis requires a multi-step procedure involving two distinct groups of software. The solid model is first exported as a Parasolid (.xmt) (Parasolid, UGS PLM Solutions, Cambridge, England) file type to preserve the closed b-rep solid properties specified in the model's creation. Other neutral file types such as IGES are not capable of supporting solid representation and are limited to the use of surface models. The parasolid file is then imported into an FE preprocessor (MSC.Patran, MSC.Software Corp, Santa Ana, CA), which serves as a user interactive front end, handling all the geometrical and modeling aspects of the FE analysis (see Figure 4.9). Once the preprocessing is completed, the load case is then solved for by a processor (MSC.Marc, MSC.Software Corp, Santa Ana, CA and ABAQUS/Explicit/Implicit, Abaqus Inc, Pawtucket, RI) and reexamined in the front end for post-processing. In order to replicate the impact experiment protocol, both the maxilla and a baseball needed to be imported into the FE software. Whereas the maxilla was created from CT scans, the solid model of the baseball was created using the closed sphere option in Rhino™ with the same dimensions as the actual ball as shown in Figure 4.10.

4.2.1 Element Selection and Mesh Generation

The fate of any good analysis using the FE method rests largely on the decisions made very early on in the modeling process. Once a mathematically sound model has been imported into the preprocessor decisions must be made regarding the appropriate selection of element type and the subsequent generation of the FE mesh. Many applications of the FE method in biomechanics have

Figure 4.9: Imported Parasolid Model of Solid Maxilla.

Figure 4.10: Solid Parasolid Model and FE Mesh of Baseball.

employed either 20-node hexahedral “brick” elements or 10-node tetrahedral elements (Hart *et al.* 1992, Horgan & Gilchrist 2003, Vollmer *et al.* 2000, Krabbel & Appel 1995 and Sujith 2000). Both element types allow for the meshing of complex geometries but the 20-node hexahedron requires the added constraint that all surfaces and curves used in the solid model be isoparametric (MSC.Patran, 2003). An isoparametric curve is defined as a Bézier curve whose distinct set of control points can be

directly computed from the equation of the surface (Shene, 2003). Since the curves used in the generation of the solid maxilla model were a generalization of Bézier curves whose control points were not evenly distributed and thus can not be directly computed from the equation of the surface, they were not isoparametric and therefore the 20-node hexahedral could not be used. As a result, any other element type that was isoparametric (wedge element, 8-node cube etc.) was also not applicable in this analysis.

In general, elements differ in their shape and their order. The shape refers to their actual geometry (be it hexahedral, wedge or shell) while the order of the element refers to the degree of polynomial used in the interpolation functions that traverse between their nodes. Theoretically, an element of higher order will converge to the analytical solution faster than a lower order element will. This is due in part to the increased number of nodes (specifically the inclusion of mid-side nodes) that allows for the improvement of the element's flexibility. A less stiff element can more easily conform to geometric changes and also absorb more strain through its deformation when loaded (Hibbitt, Karlsson Sorenson, Inc., 2002). The associated increase in conversion speed comes at the expense of increased computational time and therefore a happy medium must be selected.

Since the 20-node hexahedral element was no longer a valid choice, it was decided that tetrahedral elements be employed. The two types of tetrahedral elements available on MSC.Patran™ were the 4-node linear tetrahedral and the 10-node quadratic tetrahedral. Upon first impulse, the selection of the higher order element seems logical, however, since the aim of the FE analysis was to reproduce a transient dynamic impact, other factors must be considered. Firstly, an impact involves the contact between two or more bodies and therefore contact must somehow be detected within the FE algorithm. Some FE software packages use a Master/Slave contacting algorithm (Abaqus/Explicit, for example) in which the operator must define which is the master contacting body and which is the slave contacting body. Generally speaking the contacting body with a higher mesh density is defined as the master body and the body with lower mesh density is defined as the slave contacting body. Using a search algorithm, and the master body as a reference point, the Master/Slave algorithm sweeps the area surrounding the master body's nodes (within a predetermined radius) in search of the nodes from another body (the slave). Once a slave node is detected within this predetermined radius, contact is said to have occurred. This algorithm is a computationally inexpensive means of detecting contact between two bodies. Unfortunately, inherent within this algorithm is the tendency to detect only the corner nodes within the slave body. Should contact be initiated through a mid-side node, detection of the contact would be delayed until a corner node was present. As a result, the

deformations sustained by the slave body would be largely inaccurate. Bearing this in mind, the general protocol for FE impact simulations is to discretize with linear elements that contain no mid-side nodes solving the issue of undetected contact (Hibbitt, Karlsson Sorenson, Inc., 2002). However, as stated earlier, linear elements are overly stiff and would require a very high mesh density in order to accommodate both the complex geometry and the subsequent deformations due to the load case.

In an attempt to combat this situation, other FE software packages (MSC.Marc for instance) employ a modified Master/Slave contacting algorithm in which the operator does not arbitrarily select the contacting bodies. Instead, a simple contact table is established which specifies the number of contacting bodies. By not selecting which is master and which is slave, the search algorithm operates from all bodies and contact is no longer defined as a one-sided detection. This allows for any node to serve as the initial point of contact, be it mid-side or corner. With this new contact algorithm effectively put into place, it was decided that 10-node quadratic tetrahedral elements would populate the FE mesh. Each node on the element possessed three degrees of freedom (*dof*'s), which allowed for the translation along the *x*, *y* and *z* direction vectors.

To generate the FE mesh the auto-meshing feature on MSC.Patran™ was used with operator specified parameters. Special care must be taken in order to ensure that the elements generated during the meshing process possess acceptable geometric aspect ratios (Jacobian ratios <10) and are neither concave nor duplicated. The distortion of each element (essentially its deviation from an ideal aspect ratio) was calculated by comparing the shape of a given element in the mesh to that of an ideal element; if the shape of the element in question was identical to the ideal element it was deemed undistorted and given a value of one. Higher values meant increased distortion - this is undesirable since the ideal is 1. It is impossible to generate a mesh composed of ideal elements since the geometry, for which they are trying to conform to, is highly curved and certainly not ideal. A negative distortion value implied an element that was concave or “inside out” which is not mathematically permissible in an FE simulation.

Proper connectivity with other elements through nodal points is crucial as well since poorly connected elements will not allow the propagation of the stress wave as a consequence of the impact (Hibbitt, Karlsson Sorenson, Inc., 2002). A series of tests available to the operator allow for the verification of these element and mesh properties. Should any of these tests fail, the mesh is not valid and will not yield an accurate solution, if one at all. Iterative refining of the mesh around the areas containing troublesome elements improves their geometry until it no longer fails the verification tests.

The overall quality of the mesh is governed by its density and conformity to the solid model's geometry. To converge to the solution it is readily accepted that a higher mesh density will result in a more accurate solution. This does not at all suggest that the mesh density be constant throughout the model. Rather, regions in which high levels of strain can be anticipated should possess high levels of mesh density, such as areas of higher curvature and contact sites. Other regions, which will not be subjected to much strain, do not require a high-density mesh in order to yield an accurate solution.

The maxilla model was meshed using internal coarsening and curvature checking to govern its distribution of mesh density. These techniques ensured that the resulting mesh possessed an appropriate distribution of its density. The contact region, located at the most medial section of the maxillary incisors, also possessed a high-density mesh. This mesh density was manually created through the reseeding and splitting of the mesh in that particular area. This was important since this region would not only be the one most strained, but also, its nodes would be used in part of the contact algorithm.

The baseball itself was also meshed according to internal coarsening and curvature checking in order to yield a valid mesh. However, since the ball was to be modeled as a rigid structure the inclusion of a well-refined mesh was for the detection of contact only. The final FE model of the ball contained 2861 elements, 2861 nodes and 8583 *dof*s.

4.2.2 Convergence Testing

The overall level of mesh density required in FE modeling applications can be adequately verified by confirming a convergence in the strain energy of the simulation (Becker *et al.*, 1981). This convergence represents how effectively the numerical solution is approximating the analytical one.

A convergence test is formulated by creating a series of meshes of the same model, each one with increasing refinement, and subsequently subjecting the meshes to an identical load case. The refinement process includes the subdivision of the elements into smaller ones with the added increase of nodal points and associated *dof*s. Through a process called h-convergence, the calculated displacement of each node approaches a plateau value (albeit unknown) with each refined mesh (Hart *et al.*, 1992). If points common to each mesh are maintained and their values tracked throughout the test, the calculated displacement will converge monotonically when plotted with increasing *dof*s as its abscissa (Hart *et al.*, 1992). At some point in time, the increase in *dof* will not constitute any significant increase in displacement and therefore the mesh has converged to a value.

Five meshes were used in the formulation of the convergence test. Each mesh created was testing for validity and geometric deviance from the solid model's geometry. The size of each element dictates the number of elements required to populate a given mesh. The auto-meshing feature of MSC.Patran™ calculates the global edge length minimum. The actual element edge length is a product of this global edge length minimum and an edge length factor, which varies between 0 and 1. By altering the value of the edge length factor, the lowest possible value for an element's edge length is specified. A lower element edge length generates a finer mesh since the volume occupied by each element is smaller and therefore more of them are required to mesh the solid. The global edge length minimum for the five meshes was 0.016279 metres. The five meshes used in the convergence testing are shown in Figures 4.11 through 4.15. The material properties assigned to the convergence models were assumed to be isotropic and had a Young's modulus (E) of 20GPa and a Poisson's ratio (ν) of 0.33 (Hart *et al.* 1992). The load case common to all the convergence models consisted of a single point load of 100 Newtons applied at a single node located medially at the maxillary incisors. The boundary conditions were applied to five nodes located along the *superior* perimeter of the model with three located *anteriorly* and two located *posteriorly*. Each of these nodes was constrained for all *dof*'s in an attempt to simulate the maxilla being firmly supported by the craniofacial skeleton. The locations used for monitoring the monotonic convergence of displacement were selected based on anatomic landmarks clearly visible on the models. They consisted of both *canine fossa* and the *medial* palate. The locations of these points, the point of load application and

Figure 4.11: Mesh 1, populated by 3215 nodes, 1838 elements and 9645 *dof*'s.

Figure 4.12: Mesh 2, populated by 4135 nodes, 2377 elements and 12405 *dof*'s.

Figure 4.13: Mesh 3, populated by 7389 nodes, 4408 elements and 22167 *dof*'s.

Figure 4.14: Mesh 4, populated by 11524 nodes, 7007 elements and 34572 *dof*'s.

Figure 4.15: Mesh 5, populated by 24616 nodes, 15753 elements and 73848 *dof*'s.

the locations of the boundary conditions were kept constant throughout the test by specifying the seeding of nodes at these points through knowledge of their Cartesian co-ordinates. Once the precise spatial location of a point is known, a single node can be created at that location regardless of how dense a mesh becomes. With the locations of interest specified, the corresponding nodes are created first and the remaining mesh is built around their location.

The convergence test load case was then analyzed using ABAQUS/Implicit (ABAQUS/Implicit, Abaqus Inc, Pawtucket, RI) on a Dell Precision 650 Workstation (3.2 GHz dual-Xeon™ processor (Dell Corporation, Austin, TX). CPU times ranged from as low at 2 CPU minutes to as high as 12 CPU minutes. The convergence data was then plotted and the final mesh was deemed satisfactory for the analysis.

As a precautionary measure, a convergence test was also formulated for a series of meshes that were populated by 4-node linear tetrahedral elements. The rationale behind the use of a 4-node tetrahedral mesh was to compare the relative convergence times and degrees-of-freedom (*dof's*) required to produce a satisfactory mesh for a fundamentally different model. In a sense it was an attempt to create a “race” between the two types of models to reveal which had a more optimal design. The load case, boundary conditions and points of interest were identical to the 10-node tetrahedral convergence test. The *dof's* produced in the 4-node tetrahedral model exceeded 150,000 before any sort of convergence pattern began to emerge. Computationally, this occupied CPU times that were an order of magnitude larger than the 10-node tetrahedral model. Obviously, the 4-node tetrahedral model proved to be a very inefficient modeling technique and any other subsequent testing using this type of model was abandoned.

4.2.3 Implementation of Material Properties, Boundary Conditions and Load Case

FE modeling in biomechanics has always faced as its biggest challenge the issue of appropriately assigning material properties to the structures involved. The reason for this challenge is the simple fact that nature is largely anisotropic and non-linear. Therefore, in order to mathematically model this anisotropic, non-linear tendency, mechanical testing of bone and other biological materials must be carried out in order to reveal their true nature. Though much work has been done on long bones (please see the **Mechanical Properties of Bone Tissue** section) little or no research has been done on the measurement of the mechanical properties of craniofacial bones and/or bones with irregular shapes. FE models of the mandible circumvented this problem by assuming that the lower jaw was simply a long bone bent into a curve (Hart *et al.*, 1992). This assumption allowed for the retention of the distribution of the transversely isotropic material properties closely followed by long bones as revealed through experimentation. In general, a long bone's stiffest direction is along its long axis and this translates well in the FE model of the mandible since its directionality is somewhat intuitive (Hart *et al.*, 1992). However, in the present study, this luxury is not available since the directionality of the distribution of material properties in the maxilla is non-intuitive and any

assumptions taken from long bone experimentation would be largely arbitrary and not insightful. The unique geometry of the maxilla is unlike any other bone in the body and it simply does not reveal any tendency towards a particular axis or direction of principal strength. This, in conjunction with a lack of experimental data in the literature, led to the decision to assign isotropic material properties to the FE mesh. The rationale behind this decision was simple: rather than arbitrarily assigning directionality to an unknown system, a simpler solution incorporating much simpler assumptions carries with it less errors. This is a direct consequence of the precept of all things being equal, the simplest solution is more than likely the best solution. Keeping this in mind, the values for Young's modulus (E), Poisson's ratio (ν) and density (ρ) were taken from the literature in an average sense and assigned as 11.0 GPa, 0.3 and 1800 kg/m³ respectfully (Hart *et al.*, 1992; Fütterling *et al.*, 1998; Kober & Müller-Hannemann, 2000). Assigning it a Young's modulus whose value was an order of magnitude larger than the maxilla's artificially created the rigidity of the ball. This eliminated the possibility of any deflection occurring within the ball.

In its natural setting, the human maxilla is firmly supported within the craniofacial skeleton at its most *superior* and *posterior* points. This fixation is the result of a series of articulations between bones of the palate and the remainder of the skull. To simulate this fixation, the boundary conditions applied in the FE analysis included the constraint of all *dof*'s for a series of nodes located at the most *superior* and *posterior* regions of the model as shown in Figure 4.16.

Figure 4.16: FE Model of Maxilla with Imposed Boundary Conditions.

These boundary conditions allowed for the deformation of the maxillary incisors as well as regions in the *superior* maxilla in response to impact from a baseball traveling at a constant, initial velocity of 14 m/s. Contact detection was assigned to both bodies in the model as per the modified Master/Slave algorithm employed by MSC.Marc™ to ensure the earliest possible detection.

The baseball itself was assigned displacement constraints along the z and x axes to ensure that its impact was along the y-axis of the model analogous to the experimental situation. Though in the actual experiment, a slight backward rotation about the ball's x-axis did occur, its effect when viewed through the high speed filming was deemed negligible and thus was not included in the model.

Initially the baseball was assigned deformable material properties found in the literature (Smith *et al.*, 2000; Hendee *et. al*, 2000; Crisco *et. al*, 1997) but was later modeled as a rigid body. A value of 706.964 kg/m³ was calculated for the baseball's density. This value was derived from the definition of density as a ratio of mass to volume. With the ball's mass known (150g), its volume was calculated using the following formula where r is the baseball's radius:

$$V = \frac{4}{3}\pi r^3 \quad (7)$$

During the FE analysis, the ball experienced unrealistic deformations not present in the actual experiment when the literature values were assigned. Since no similar experiments values were available to validate the results, the deformable ball model was abandoned in favor of a rigid ball. When viewed through high speed filming, the ball's deformation in comparison to the maxilla's is negligible and therefore the application of a rigid ball within the model was said to be adequate.

The load case for the impact was modeled as a transient dynamic simulation. This involved the assignment of time based variables inherent in the iterative solution process. The duration of the analysis was set to 0.1 seconds long. This included the time required for the baseball to traverse the initial pre-impact distance of 5mm, the impact itself and the subsequent post-impact rebound. The time stepping required for the dynamic simulation was an automatic time-stepping scheme employed by MSC.Marc™. This scheme structures the size of its time steps based on the time required for a stress wave to propagate through the structure. This time span is a function of several parameters including the dimension of the model's smallest element. In general, the smaller the element, the shorter the time step required yielding a higher number of iterations for the solution. Since the outcome of the solution is highly sensitive to the changes made in these parameters, MSC.Marc™ automatically and dynamically assigned them.

Transient dynamic simulations can be solved by either implicit or explicit solution schemes. For the case of large systems with extreme velocities and accelerations, explicit code is preferred since the implicit routine involves the inversion of matrices every iteration that is computationally expensive. Special care must be exercised when employing an explicit scheme since convergence is not always guaranteed unless all parameters are well understood and judiciously selected. The maxilla impact simulation did not require the use of an explicit routine since the magnitude of the velocities and the model itself were small. Figure 4.17 shows the FE model of the maxilla and the ball in their pre-impact positions.

Figure 4.17: FE Model of Maxilla and Ball in Pre-Impact Position.

4.2.4 Solution and Validation of the FE Model

The solution of the FE model was carried out using the MSC.Marc™ solver through the MSC.Patran™ (MSC.Marc & MSC.Patran, MSC.Software Corp, Santa Ana, CA) preprocessor on a Dell Precision 650 Workstation (3.2 GHz dual-Xeon™ processor) (Dell Corporation, Austin, TX). Included within the solution were calculated values for nodal displacements, nodal stress invariants, reaction forces at the constrained nodes and the stress/strain tensor components.

In the experimental portion of this study, cadavers were impacted and fractures were induced in the maxillofacial region. Experimental data was recorded by way of strain gauges that, in theory, measure only the principal (or normal) stress at the surface of the specimen they are bonded to. Peak

strain was assumed to have occurred at the moment of fracture. This assumption was made because the implementation of a failure model to simulate fracturing was not possible due to a lack of knowledge regarding the directional nature of the maxillary bones and their subsequent thresholds for stress/strain. All failure models in MSC.Patran™ required knowledge of these properties *a priori* and therefore implementation of said models was not possible. Therefore, a comparison of the peak principal strains (assumed to be at the moment of fracture) measured from each rosette during the experimental testing, to the calculated values of peak principal strain derived from the model, served as the validation of the FE analysis (Korioth, 1992). Differences between the two fields were expressed as relative error percentages. Since the locations of both the gauge rosettes and nodal points of interest were based on clear physiological landmarks there was little uncertainty regarding the coincidence of their locations. For the sake of skepticism and redundancy though, these locations were visually inspected to confirm geometric compatibility.

Voltage values outputted from the strain gauge amplifier and the load cell amplifier/conditioner included amplification constants and factors which required removal in order to reveal the true output values of the instrumentation (be it microstrain or Newtons). The sensitivity of the strain gauge amplifier as given by the manufacturer's manual is described by the following equation:

$$e = E \cdot \frac{n}{4} \cdot GF \cdot \varepsilon \quad (8)$$

where e is the operator defined sensitivity in millivolts, E is the operator defined excitation voltage, n is the number of active gauges within the Wheatstone bridge, GF is the gauge factor and ε is the actual strain output of the gauge. The sensitivity of the amplifier was set to 5mV; the excitation voltage at 5V and the number of active gauges with a 2.12 gauge factor within the bridge was one. Re-arranging equation (8) and solving for ε yields a strain of 1886 $\mu\varepsilon$ that a 5mV, operator defined sensitivity will produce. All other output values from the amplifier can be expressed as a percentage of the excitation voltage and therefore, a percentage of the 1886 $\mu\varepsilon$ subject to the same sensitivity.

Retrieving the absolute maximum voltage values from the strain gauge data using custom written MATLAB programs and subsequently solving for their percentage of amplifier-sensitivity based strain gave the peak strain values from the impact. Using these values, the principal strain magnitudes were calculated according to the following formula:

$$\varepsilon = \frac{1}{2}(\varepsilon_0 + \varepsilon_{90}) \pm \frac{1}{2}\sqrt{[(\varepsilon_0 - \varepsilon_{90})^2 + (2\varepsilon_{45} - \varepsilon_0 - \varepsilon_{90})^2]} \quad (9)^2$$

where ε_0 is the measured strain value at the selected reference gauge assigned by the manufacturer as the leftmost gauge (indicated by a number 1). The remaining variables ε_{45} and ε_{90} correspond to the strain values measured by the other two gauges present in the rosette at 45 and 90 degree orientations measured clockwise from the reference gauge.

The principal angle associated with the principal strain was calculated using the following formula:

$$\tan 2\theta = \frac{(\varepsilon_0 - 2\varepsilon_{45} + \varepsilon_{90})}{(\varepsilon_0 - \varepsilon_{90})} \quad (10)^3$$

The output from the load cell amplifier/conditioner arrived with units of voltage. In order to obtain values for forces in Newtons and moments in Newton metres, a custom written MATLAB program pre-multiplied all the signals with the following calibration matrix:

$$\begin{bmatrix} 451.7 & 37.3 & -2.5 & -2.5 & 1.5 & -2.9 \\ 26.6 & 453.1 & -1.0 & -4.3 & -0.5 & 0.1 \\ 11.7 & 10.6 & 378.2 & -3.6 & -4.2 & -11.4 \\ -1.1 & -16.4 & -0.2 & 5.6 & -0.1 & 0.0 \\ 16.2 & 1.5 & 0.0 & -0.1 & 5.6 & -0.2 \\ 0.2 & 0.0 & 0.0 & 0.0 & 0.0 & 8.4 \end{bmatrix} \quad (11)^4$$

the units of the above matrix are Newtons/Volt for the first three rows corresponding to the force signals and Newton Metres/Volts for the last three rows corresponding to the moment signals.

As a secondary means of validation, the calculated maximum displacement of the *central* incisors from their natural position in the model (analogous to a luxation injury) was compared to the displacement values seen and treated by Dr. David Kenny and Dr. Michael Casas on patients' suffering similar injuries.

² Taken from Beer & Johnston, 1992, pp. 345.

³ Taken from Beer & Johnston, 1992, pp. 345.

⁴ Taken from Bertec Corporation, 2001, pp. 1.

Chapter 5

5 RESULTS

The results from the procedures outlined in **Chapter 3: Experimental Methodology** and **Chapter 4: Modeling Methodology** are presented herein. All attempts were made to preserve the sequential integrity of the outlined procedures although some deviation has occurred for the sake of clarity. Preliminary experimentation and developmental exercises will be presented first, followed by the impact testing on cadavers and ultimately the transient dynamic FE modeling of the experimental proceedings.

5.1 Preliminary Experimentation and Developmental Exercises

Preliminary exercises involving the establishment and qualitative verification of the experimental protocol, namely the oxidized beam and bovine rib exercises, will not be included since a quantitative analysis was not performed. The aim of those exercises was the development of operator confidence and a verification of the outlined policies and procedures involved in the application of electrical resistance strain gauging for the experimental measurement of strain found in the literature. Several adaptations to the existing protocol were made due to the insights gained from the implementation of these developmental exercises and therefore their inclusion within the text was logical. A qualitative discussion of the insights gleaned and their relevance to the adapted protocol is included in Section 6.1.

5.1.1 Mock Experiment Using a Plastic Model Skull

The duration of data acquisition for the mock experiment was set at 5 seconds giving a total of 33750 samples of data taken at 6750Hz. At this rate, each data sample required 1.48×10^{-4} seconds and therefore a clear and definite spike would occur in the data at the moment of impact. The data acquisition program plotted the voltage of the attenuated signal versus the number of samples taken during the experiment (see Figures 5.1 through 5.9). Due to a corruption of the soldering terminal, data from the first direction of the strain gauge was unavailable during the impact experiment and thus principal strains could not be calculated.

Figure 5.1: Mock Experiment Strain Gauge Data: Direction 1.

Figure 5.2: Mock Experiment Strain Gauge Data: Direction 2.

Figure 5.3: Mock Experiment Strain Gauge Data: Direction 3.

Figure 5.4: Mock Experiment Load Cell Data: Force Along x-axis.

Figure 5.5: Mock Experiment Load Cell Data: Force Along y-axis.

Figure 5.6: Mock Experiment Load Cell Data: Force Along z -axis.

Figure 5.7: Mock Experiment Load Cell Data: Moment About x -axis.

Figure 5.8: Mock Experiment Load Cell Data: Moment About y-axis.

Figure 5.9: Mock Experiment Load Cell Data: Moment About z-axis.

The ball impacted the skull medially at the maxillary incisors at a velocity of 24.14m/s (54mph) as measured by the Speedchek doppler radar gun. This resulted in a classic LeFort III fracture (see Figure 5.10) otherwise known as a craniofacial disjunction. As a consequence of this catastrophic failure of the skull, several of the lead wires were detached from the gauge and therefore a premature ending of data acquisition occurred for those particular channels. The collection of data shows a high peak-signal-to-peak-noise ratio, which was concluded through a qualitative observation of the peak signal compared to the peak noise. The result was an abundantly clear and definite spike in the signal corresponding to the precise moment of impact. A quantitative analysis of the peak-signal-to-peak-noise ratio could have conducted but was deemed un-necessary after the qualitative observation was so pronounced. Maximum values for strain and load cell voltage are included in Table 5.1.

Figure 5.10: Plastic Skull Post-Impact Demonstrating a LeFort III Fracture.

The data acquisition system proved to be reliable in its ability to accurately capture the impact profile. Though some adjustments to the amplification value of the load cell signal conditioner and a reduction of the overall impact velocity, circuit saturation and clipping was easily avoided for the cadaver testing.

Measured Quantity	Maximum Voltage	Minimum Voltage
Strain Gauge Direction 1	N/A	N/A
Strain Gauge Direction 2	10.0	-10.0
Strain Gauge Direction 3	7.225	-10.0
Force Along x -axis	6.0278	-7.1414
Force Along y -axis	7.2189	-7.0153
Force Along z -axis	6.3867	-7.1362
Moment About x -axis	7.2406	-7.1438
Moment About y -axis	7.1789	-7.1398
Moment About z -axis	6.8118	-7.1380

Table 5.1: Maximum and Minimum Voltage Values from Mock Experiment.

5.2 Experimental Impact to Cadaver Heads

5.2.1 Qualitative Results of Impact Experiment

As expected, the injuries resulting from the impact experiment were acute and strongly localized at the medial maxilla as well as its accompanying incisors. Although the severity of the injuries varied from subject to subject, the type of injury (avulsion, root fracture etc.) remained constant throughout the study verifying the validity of the experimental protocol. Dr. David Kenny and Dr. Michael Casas diagnosed the injuries induced in the impact experiment according to standardized WHO procedures. In order to illustrate the range and types of injuries observed in the experimental study, all specimens impacted will be presented and their injuries qualitatively described.

5.2.1.1 Cadaver Specimen #5164

Specimen number 5164 was impacted medially at the central maxillary incisors with no soft tissue present. The following figures (5.11 & 5.12) illustrate the location of the impact and the post-impact trauma.

Figure 5.11: Frontal and Lateral Views of Impact to Specimen #5164.

Figure 5.12: Frontal and *Inferior* Views of Specimen #5164, Post-Impact.

The impact occurred medially at the maxillary incisors resulting in several injuries. The most prominent injury was a LeFort I fracture involving the 11, 12, 13 and 14 maxillary incisors. In addition to this LeFort I fracture, the 11, 12, 21, 23 incisors incurred cervical root fractures. The trauma resulting from this impact left no intact maxillary incisors.

5.2.1.2 Cadaver Specimen #5180

The soft tissue for specimen #5180 was reflected to allow for the placement of the strain gauge assemblies and therefore was intact for the duration of the impact experiment. The impact was again located medially at the maxillary incisors but slightly more inferior than the impact site for specimen #5164. Post-impact trauma is illustrated in the following figures (5.13 & 5.14).

Figure 5.13: Frontal and Left *Iso-lateral* Views of Specimen #5180, Post-Impact.

Figure 5.14: Right *Iso-lateral* Views of Specimen #5180, Post-Impact.

The injuries incurred after the impact included a 12 mid-crown fracture, a 11, 21 mid-crown fracture coupled with a lateral luxation and finally a 22 cervical one-third fracture.

5.2.1.3 Cadaver Specimen #5186

The soft tissue was also reflected and kept intact for the impact of specimen #5186. Once again, the impact site was located at the medial region of the maxillary incisors at the height of the upper labia. This resulted in a 12 lateral luxation, 11, 21 avulsion and an alveolar fracture from 11 to distal 12. The right, canine fossa strain gauge was also damaged during the experiment due to the propagation of the alveolar fracture. These injuries and the impact site are illustrated in the following figures (5.15 & 5.16).

Figure 5.15: Frontal Views of Impact and Post-Impact Trauma for Specimen #5186.

Figure 5.16: Right *Iso-lateral* and *Inferior* Views of Specimen #5186, Post-Impact.

5.2.1.4 Cadaver Specimen #5199

The following figures (5.17 & 5.18) illustrate the injuries incurred by specimen #5199. These injuries include 11 and 12 avulsions, a 22 subluxation, 12 and 13 lateral luxations and finally an alveolar fracture starting at 22 to distal 13. The ball impacted at the maxillary incisors, as intended, and was met by both soft and hard tissues.

Figure 5.17: Frontal and Left *Iso-lateral* Views of Specimen #5199, Post-Impact.

Figure 5.18: Right *Iso-lateral* and *Inferior* Views of Specimen #5199, Post-Impact.

5.2.1.5 Cadaver Specimen #5212

The soft tissues of specimen #5212 were left intact for the impact experiment. The ball impacted both hard and soft tissue at the maxillary incisors and produced the following injuries: a 23 lateral luxation, a series of 11,21, 22 avulsions and a labial blowout fracture. The extent and severity of the injuries as a result of the impact is depicted in the following figures (5.19 & 5.20).

Figure 5.19: Frontal and Left *Iso-lateral* Views of Specimen #5212, Post-Impact.

Figure 5.20: Right *Iso-lateral* and *Inferior* Views of Specimen #5199, Post-Impact.

5.2.1.6 Cadaver Specimen #5247

All soft tissues surrounding the maxilla, including the nose, were removed for specimen #5247. This resulted in a hard tissue-only impact at the medial region of the maxillary incisors. Using the figures 5.21 & 5.22 as a visual aid, the following injuries were observed: an alveolar fracture distal 13 to the zygoma, a series of 11, 12 and 13 avulsions, a 21 cervical root fracture, a 22 lateral luxation and a 21, 22 labial plate blowout. As a result of the propagation of the alveolar fracture, the patient's right, canine fossa strain gauge assembly was torn off and damaged. Similarly, the patient's left, canine fossa strain gauge assembly was also damaged during the impact.

Figure 5.21: Frontal and Right *Iso-lateral* Views of Specimen #5247, Post-Impact.

Figure 5.22: *Lateral* and *Inferior* Views of Specimen #5247, Post Impact.

5.2.2 Quantitative Results of Impact Experiment

The six cadaver heads were all successfully impacted as set forth by the experimental protocol. However, factors such as accumulated moisture and signal degradation led to the unsuccessful acquisition of data from random strain gauge channels. In one impact (Cadaver #5199) all nine strain gauge channels successfully acquired data. It was then decided that taking an average value of all the available data for each channel would be the best approach to incorporate all the data from all the heads. On average, each channel had 4 independent impact values for which the mean was determined. Data from the load cell was successfully collected in all of the impact trials.

5.2.2.1 Cadaver Specimen #5199

For the sake of brevity, the complete data set including all strain gauge and load cell data for only one cadaver head (#5199) is presented within the body of the text (Figures 5.23 - 5.37) along with the averaged values. For the sake of completeness, the data set chosen represents the fullest set of acquired experimental values. The remaining data will be plotted and shown in **Appendix B**.

Figure 5.23: Cadaver Experiment Strain Gauge Data: Gauge 1 Direction 1.

Figure 5.24: Cadaver Experiment Strain Gauge Data: Gauge 1 Direction 2.

Figure 5.25: Cadaver Experiment Strain Gauge Data: Gauge 1 Direction 3.

Figure 5.26: Cadaver Experiment Strain Gauge Data: Gauge 2 Direction 1.

Figure 5.27: Cadaver Experiment Strain Gauge Data: Gauge 2 Direction 2.

Figure 5.28: Cadaver Experiment Strain Gauge Data: Gauge 2 Direction 3.

Figure 5.29: Cadaver Experiment Strain Gauge Data: Gauge 3 Direction 1.

Figure 5.30: Cadaver Experiment Strain Gauge Data: Gauge 3 Direction 2.

Figure 5.31: Cadaver Experiment Strain Gauge Data: Gauge 3 Direction 3.

Figure 5.32: Cadaver Experiment Load Cell Data: Force Along x -axis.

Figure 5.33: Cadaver Experiment Load Cell Data: Force Along y -axis.

Figure 5.34: Cadaver Experiment Load Cell Data: Force Along z -axis.

Figure 5.35: Cadaver Experiment Load Cell Data: Moment About x -axis.

Figure 5.36: Cadaver Experiment Load Cell Data: Moment About y -axis.

Figure 5.37: Cadaver Experiment Load Cell Data: Moment About z -axis.

The data acquisition program calculated the principal strain values in two dimensions using equations outlined in Section 4.2.4. The results of these calculations for the specified gauge locations on specimen 5199 are shown in table 5.2.

The surface temperature of both the cadaver and the strain gauge assemblies was stable at 19.0°C that is within the acceptable range as specified by the manufacturer and therefore temperature compensation was not required for the outputted values.

Gauge Location	Principal Strain Max. ($\mu\epsilon$)	Principal Strain Min. ($\mu\epsilon$)	Direction (°)
Gauge 1 (S.R. Fossa) ⁵	4636	544	-63
Gauge 2 (M. Palate) ⁶	2825	1237	-52
Gauge 3 (S.L. Fossa) ⁷	3223	208	-23

Table 5.2: Calculated Principal Strain values for Specimen 5199.

5.2.3 Relevant Values for the Remaining Specimens and Average Peak Principal Strains

Values from the remaining specimens were averaged by first removing local extrema values and then the arithmetic mean was calculated. The removal of the high and low extreme values was done in order to eliminate any possible values that may skew the measured results in one direction or another. Similar to specimen #5199, surface temperature values for the remaining samples were well within the manufacturer's specifications and as a result it was assumed that no temperature drift occurred and was thus not compensated for. Values for craniofacial and palate surface temperatures along with the averaged principal strain values are presented in tables 5.3 and 5.4.

Raw data from the load cell were also averaged as voltage values and then converted to values of force (N) and moment (Nm) employing the transformation matrix and the signal conditioner's amplification values as outlined in Section 4.2.4. Only the peak values of the voltage were used in the calculation and once again, extrema were not included in the calculation of the arithmetic mean. The results of these calculations are shown in tables 5.5 and 5.6.

Specimen Number	Facial Surface Temp. (°C)	Palatal Surface Temp. (°C)
5164	19.5	19.0
5186	19.5	19.5
5180	18.0	17.5
5212	19.0	18.5
5247	20.5	19.0

Table 5.3: Facial and Palatal Surface Temperature Values for Impact Experiments.

⁵ Specimen's right *canine fossa*.

⁶ Specimen's *medial* palate.

⁷ Specimen's left *canine fossa*.

Gauge Location	Averaged Principal Strain Max. ($\mu\epsilon$)	Averaged Principal Strain Min. ($\mu\epsilon$)	Direction ($^\circ$)
Gauge 1 (S.R. Fossa)	4243	369	-70
Gauge 2 (M. Palate)	2371	1392	-41
Gauge 3 (S.L. Fossa)	3819	390	-19

Table 5.4: Averaged Principal Strain Values for all Impact Experiments.

Force/Moment	5164 (V)	5186 (V)	5180 (V)	5199 (V)	5212 (V)	5247 (V)	Average (V)	Actual Value
Fx	0.986	0.445	1.78	1.24	1.15	1.21	1.15	102.7 N
Fy	0.394	0.174	0.989	0.871	1.54	3.23	0.949	83.6 N
Fz	1.13	0.524	1.55	0.865	5.88	5.82	2.341	72.9 N
Mx	5.88	1.93	7.21	3.76	5.33	7.21	5.545	6.9 Nm
My	3.36	1.77	4.99	2.57	2.95	3.62	3.125	3.2 Nm
Mz	1.61	1.20	1.19	1.94	2.41	1.99	1.685	3.2 Nm

Table 5.5: Averaged Maximum Values for Load Cell Data.

Force/Moment	5164 (V)	5186 (V)	5180 (V)	5199 (V)	5212 (V)	5247 (V)	Average (V)	Actual Value
Fx	-0.37	-0.21	-1.1	-0.63	-6.31	-5.62	1.93	-54.56 N
Fy	-2.38	-1.09	-2.15	-1.77	-1.606	-1.45	1.74	-146.69 N
Fz	-1.39	-0.508	-4.05	-1.84	-1.54	-3.06	1.96	-129.39 N
Mx	-7.05	-4.25	-7.05	-7.05	-7.04	-7.04	7.045	-10.40 Nm
My	-3.41	-2.52	-7.01	-6.97	-4.47	-7.00	6.37	-6.05 Nm
Mz	-3.01	-1.05	-1.82	-3.6	-2.41	-1.92	2.29	-6.01 Nm

Table 5.6: Averaged Minimum Values for Load Cell Data.

5.3 FE Modeling and Results

The results of the FE analysis were divided into two categories, those dealing with the validation of the mesh and model itself in a static load case, and those dealing with the transient dynamic simulation of the impact experiment.

5.3.1 Mesh Validation and Convergence Testing via a Static Load Case

The validation of the FE model, specifically the populated mesh, was accomplished via a number of specific tests designed to confirm the overall conformity of the mesh both mathematically and geometrically. These tests include a check for any duplicate elements and nodes, element connectivity, aspect ratios (Jacobian) and an overall fit-error to the actual imported solid model's geometry. These tests were conducted for each of the meshes and used in conjunction with the convergence testing, in a static load case. All models included within the convergence testing were fully equivalent, connective and free of any duplicate elements and nodes.

For the actual convergence testing, a single, 100N load was applied medially along the y-axis of the model on a single node located just *superior* to the central maxillary incisors. The convergence test focused on the displacement, in millimeters, for three points of interest (nodes A, B and C) after the applied loading. These points corresponded to the same physiological landmarks chosen for the location of the strain gauge assemblies during the impact experimentation; namely the left and right canine fossa and the medial palate (node numbers 19560, 12605 and 22396 respectfully). The geometric locations of these landmarks/nodes are presented in Table 5.7 relative to the solid model's centre of mass. A clear monotonic convergence towards a distinct numerical value for each node in question was the result of the increasing mesh density. This objectively demonstrates the fact that the densest meshes are producing accurate results. For meshes 4 and 5, the difference between the calculated displacements for nodes A, B and C were only 2 to 3% while the associated increase in *dof* is well over 100%. As a consequence of this slight change in displacement versus a large increase in associated *dof*, we can confidently assume that the final and most refined mesh is yielding accurate results at the cost of 73,000 *dof*. Details regarding the results of the mesh validity testing are tabulated in Table 5.8 and the results of the convergence testing are shown in Table 5.9 and plotted in Figures 5.38 through 5.40.

Point of Interest	x co-ordinate (m)	y co-ordinate (m)	z co-ordinate (m)	Node Name/Number
Right Fossa	0.002782075	-0.022523304	-0.0081854779	A/1
Medial Palate	0.0025483654	-0.018070003	-0.0081854779	B/2
Left	0.020321423	-0.012209106	0.00087702234	C/3

Table 5.7: Geometric Locations of Convergence Points Relative to the Applied Load Location.

Model Name	Global Edge Length (m)	Edge Length Factor (m)	Edge Length Minimum (m)	Geometry Fit Error (m)	Maximum Jacobian Ratio	Maximum Jacobian Zero
Converge1	0.016279	0.4	0.0065116	0.00145	194	5.48×10^{-7}
Converge2	0.016279	0.3	0.0048837	1.38×10^{-3}	71	2.93×10^{-7}
Converge3	0.016279	0.2	0.0032558	1.04×10^{-3}	8.4	1.61×10^{-7}
Converge4	0.016279	0.15	0.0024419	1.01×10^{-3}	5.9	1.58×10^{-7}
Converge5	0.016279	0.125	0.0020035	1.01×10^{-3}	5.4	1.53×10^{-7}

Table 5.8: Mesh Validity Tests Performed on all Meshes Used in the Convergence Test.

Model Name	% Change in Displacement		
	Node A	Node B	Node C
Converge1	N/A	N/A	N/A
Converge2	12.68	37.47	2.58
Converge3	15.27	14.46	22.08
Converge4	2.76	7.56	4.79
Converge5	1.67	3.05	1.49

Table 5.9: Percent Change in Calculated Displacement for Points of Interest Illustrating a Numerical Convergence to a Unique Solution.

Figure 5.38: Convergence Plot Showing the Change in Calculated Displacement for Node A.

Figure 5.39: Convergence Plot Showing the Change in Calculated Displacement for Node B.

Figure 5.40: Convergence Plot Showing the Change in Calculated Displacement for Node C.

5.3.2 Comparison Between Experimental and Calculated Strains from a Transient Dynamic Simulation

A comparison between the calculated maximum principal strains in two dimensions from the transient dynamic simulation and the averaged maximum principal strains (again in two dimensions) derived from the experimental data is shown in table 5.10. Principal strains are shown for the three chosen physiological landmark locations and the differences between the two sets of values is expressed as a percentage of relative error. For the simulated impact experiment, the peak of the impact occurred on the 48th iteration of the solution and this translated to a time of 5.077×10^{-4} seconds.

Anatomical Location Within Cadaver and Model	Measured Maximum Principal Strain in 2-D (ϵ)	Calculated Maximum Principal Strain in 2-D (ϵ)	Percentage of Measured Value (%)	Relative Error (%)
R. Canine Fossa	0.004243	0.004374	103.07	3.07
Medial Palate	0.002371	0.0021013	88.62	11.37
L. Canine Fossa	0.003819	0.0038828	101.67	1.67

Table 5.10: Relative Differences Between Observed and Calculated Maximum Principal Strain Values in Two Dimensions on the Cortical Surface of the Human Maxilla.

If compared to the estimated value of 0.44 milliseconds for impact duration found in the literature, we see a relative difference of 15.38% for overall system response time (Crisco *et al.*, 1997).

The maximum displacement of any point on the maxillary central incisors also occurred at the moment of peak impact. This displacement occurred at the most inferior nodes of the maxillary incisors and equaled a value of 2.7mm along the anterior-posterior axis of the maxilla in the posterior direction.

The relative differences for the magnitudes of maximum principal strain ranged from 1.67% to 11.37%. The largest difference occurred within the comparison of values calculated and gathered for the medial palate or strain gauge assembly 2/node B. In general, the overall average difference in values was approximately ~5%.

Figures 5.41 through 5.45 show some of the deformations calculated in the FE simulation both in wire mesh and contour plot views.

Figure 5.41: Inferior View of Maxillary Deformation During a Simulated Impact with a Rigid Ball.

Figure 5.42: Inferior, Wire Frame View of Maxillary Deformation During a Simulated Impact with a Rigid Ball.

Figure 5.43: Lateral View of a Contour Plot of Simulated Impact, with Rigid Ball Visible.

Figure 5.44: Lateral View of a Contour Plot of Simulated Impact, with rigid Ball Hidden.

Figure 5.45: Inferior, Wire Frame View of Maxillary Deformation at Maximum Deflection.

Chapter 6

6 DISCUSSION

6.1 Preliminary Exercises and Experimentation

6.1.1 Application of Strain Gauge Assemblies to the Surfaces of an Oxidized Steel Beam and an Embalmed Bovine Rib

The experimental measurement of strain from the surfaces of an oxidized steel beam and bovine rib proved to be very useful in developing a working protocol which could then be extended to the application of strain gauge assemblies to embalmed human maxillæ. The application of strain gauge assemblies to the oxidized steel beam was much less labour intensive and complicated than for the case of the bovine rib. The main reason for this simpler application was the flat, uniform surface present on the oxidized steel beam as opposed to the sharply convex surface of the embalmed bovine rib. This allowed for much a much easier preparation of the surface and subsequent bonding of the strain gauge assemblies since an even amount of thumb pressure could be distributed onto the backing of the silicon-carbide paper and strain gauges.

Furthermore, since the embalmed bovine rib was a “wet” surface, with constant re-wetting occurring due to the seeping out of embalming fluid from the porous interior, the sealing of the surface with a thin layer of methyl-2-cyanoacrylate and its associated catalyst was necessary to prevent leaking of fluid onto the strain gauge assembly itself. Time was also a factor since the oxidized steel beam could be abraded and surface treated without any danger of over-drying, as would be the case for the embalmed bovine rib.

6.1.2 Mock Impact Experiment Using a Plastic Skull

The differences between the proposed experimental protocol and the actual procedure implemented within the cadaveric experimentations confirmed the notion that the execution of a “dry run” experiment was prudent and not an act of redundancy. The alterations made to the impact velocity and the accuracy control were significant in that a reduction in impact speed and a more precise impact location increased the probability of inducing maxilla-specific injuries as opposed to complex and coupled fractures which frequently occur at higher impact energies (Richardson, 2000).

The inclusion of a secondary, frontal high-speed camera was also the result of the mock experiment. It was decided that a second camera was necessary to document the precise location of the impact in a medial-lateral sense, since an isolated lateral camera could only reveal details in a superior-inferior sense. By analyzing both sets of images, a conclusion could be drawn, with confidence, that the site of impact was in fact the desired location, namely the medial incisors.

Furthermore, since high-speed impact testing of this nature, involving a freely moving impactor as opposed to an impactor constrained in some manner (be it rotationally or translationally), has not been used before in the application of impact tolerances of the human maxilla, it was therefore necessary to document as accurately as possible the details of the impact location as it happened, since this knowledge was not available with 100% confidence a priori.

The results of the mock experiment revealed the details required and the level of control necessary to ensure the successful completion of the actual cadaver impact experiments. The application of the strain gauge assemblies onto a geometry identical to that of the actual experimental sample further increased the technical proficiency of the experimenter and thus confirmed the overall effectiveness of the application procedure as seen and modified from the literature (Szivek and Gharpuray, 2000).

6.2 Impact Experiments with Embalmed Cadaver Skulls

Earlier investigations have revealed the impact tolerances of the various bones of the craniofacial complex. Localized trauma was successfully produced, and the resulting bone-specific fractures have created a large database on the topic. However, the nature of the impacts has been highly mechanical, involving high mass-low velocity impactors whose trajectories were heavily constrained (Hodgson *et al.*, 1967; Schneider and Nahum, 1974; Nyquist *et al.*, 1986; Welbourne *et al.*, 1989).

Furthermore, a quantification of the level of strain induced in such an impact, and the ultimate strains resulting in the subsequent fracturing of the bone has been largely ignored. Instead, the focus has been on the overall forces required to fracture facial bones. The study conducted herein focused on both the overall forces generated by the impact, as well as the induced strains at specific locations on the maxilla. The use of an unconstrained, low mass-high velocity impactor was intended to mimic the types of impact situations which occur in everyday settings. It is not likely that a slow-moving steel rod, traveling along a guided track will impact an unsuspecting person precisely in the anterior maxilla. On the other hand, the dental trauma unit at The Hospital for Sick Children in Toronto, Canada routinely treats patients who have suffered dental trauma as a result of an errant baseball or hockey puck.

Figures 6.1 and 6.2 show a direct comparison of the impact experiment results and the trauma sustained by actual patients involved in similar high-velocity, low-mass impacts with baseballs. Dr. David Kenny and Dr. Michael Casas at The Hospital for Sick Children in Toronto, Canada treated the patients depicted in these figures. Upon close examination of the figures we see a very striking resemblance between the compared scenarios.

Figure 6.1: A Direct Comparison of the Resulting Crown Fractures from the Cadaver Impact Experiment and a Patient Treated at The Hospital for Sick Children in Toronto, Canada.

Figure 6.2: A Direct Comparison of the Resulting LeFort Style and Alveolar Fractures from the Cadaver Experiment and a Patient Treated at The Hospital for Sick Children in Toronto, Canada.

The first figure shows a series of mid-crown fractures common to both maxillæ. Note the highly localized injury patterns. The extent of the trauma sustained by both cadaver and patient resides almost exclusively at the four anterior maxillary incisors. The second figure depicts injuries that are much more severe. Here we see avulsions coupled with LeFort I and alveolar fractures. The precise location, radius and extent of the trauma corresponds exactly with what has been reported in the literature as the most common and typical variety as a result of blunt impact to the anterior maxilla (Andreasen, 1994; Bijur *et al.*, 1995; Rupp, 2002). The qualitative comparison of the impacts produced as a result of the impact experimentation with those of patients treated in the emergency room serves as a validation that the impact scenario, albeit still highly controlled, captured a typical subject sustaining a typical injury.

The experimental measurement of strain from the exposed bony surfaces of the maxilla involved several restrictions that were not present in the application of strain gauge assemblies to the oxidized steel beam and the embalmed bovine ribs. First, the cadaver maxillæ were stored in a cold environment prior to surface preparation and therefore the polymerization process of the methyl-2-cyanoacrylate adhesive, which uses heat as its catalyst, required up to three times longer to fully cure compared to a room temperature setting. This, combined with the highly curved geometry of the maxilla, created complications throughout the application process. It was found that the corners of the strain gauge assemblies were initially not conforming to the surface curvature well and therefore not bonding to the bone when using a single thumb as a means of pressure and heat. Using two thumbs or portions of the experimenter's palm to provide a larger surface area of heat and pressure combated this problem. Once bonded into place the lead wires of the gauges were soldered into place and checked for resistance. It was observed that the strain gauge assemblies were very sensitive to leaks from the surrounding moist environment leading to erroneous resistance values and ultimately false readings of strain or lack thereof. In prior research it has been shown that silicone rubber, brushed over the surface of the bonding terminals is adequate for isolating the strain gauge lead wires (Szivek and Gharpuray, 2000). In order to further isolate the terminals, a sheet of rubber dental dam was bonded in a flap-like fashion above the entire assembly. The dental dam served as an impermeable membrane thus preventing any further leaks from the surrounding moist medium.

In general, the strains experienced by the cadaver maxillæ were on the order of $4000\mu\epsilon$ for the canine fossæ and $2300\mu\epsilon$ for the medial palate. This result is intuitive since the medial palate would be largely exposed to a compressive load and would be shielded from large strains by the remaining tissues. The strains experienced by the canine fossæ would be induced through a dominant

bending force on the superior maxilla's exterior surface. Since these points are located at the outermost regions of the maxilla and thus are the closest to the impact site, it is no surprise that the magnitudes of their strains is significantly higher than those found at the medial palate.

The load cell data also revealed very intuitive results. The orientation of the load cell was such that its x -axis was along the left-right direction of the cadaver; its y -axis was along the anterior-posterior direction and the z -axis along the superior-inferior direction. As a result of the impact to the anterior portion of the cadaver, the largest force should be experienced along the y -axis of the load cell and the largest moment should be about the load cell's x -axis. This in fact was the case as the results of the data acquisition clearly indicate. Curiously, the moment about the load cell's z -axis was also quite high in comparison to the other values. This implies a slight torquing of the cadaver(s) as a result of the baseball impact and could be the result of a non-uniform distribution of mass within head. Since most of the mass was concentrated towards the posterior and inferior portions of the head, a slight deviation from a precisely medial impact would be exploited by this imbalance and thus be manifested as an increase in the moment about the load cell's z -axis.

The strain gauge and load cell data outputted from the transducers was, in general, successfully recorded by the data acquisition system. However, some channels from the strain gauge assemblies did not successfully capture data. This was due to the development of condensation on the gauges and lead wires when stored in the cold environment of the morgue. An attempt to remove all condensation with cotton gauze and swabs was successful in some cases but led to a further corruption of the terminals in others. Ultimately, the result was the unsuccessful reading of data from specific channels, which differed (randomly) from experiment to experiment. In order to employ all available data from all experiments, it was decided to assemble data from each channel and calculate the arithmetic mean of all the working channels and thus develop values for validation purposes in an average sense.

Data from all working channels (both strain gauge and load cell) behaved similarly in the capturing of data both pre and during the impact. In general, a sharp definite spike in the data corresponding to the impact itself followed a region of low (essentially zero) signal. The post-impact data samples differed largely from experiment to experiment. In most cases, the amplitude of the incoming signal dropped back down to the same values as the pre-impact sampling. This corresponds to the scenario where the gauges are strained to a peak within their elastic range and returned to their original undeformed state without any permanent deformation. In other cases, the post impact signal did not return to its original configuration as per the pre-impact signals. Instead, the post-impact

signals dropped to a value higher than those observed pre-impact. This corresponds to the scenario where the gauges are strained to a peak that is beyond their elastic range and therefore permanent plastic deformation of both gauge backing and foil will occur. After the load is removed, the gauge remains in a deformed state and therefore will have a slight increase in the resistance across its foil giving a non-zero signal. In the most extreme deformations, the gauge itself (or the attached lead wires) is strained beyond tolerance and fails, causing an instant termination of the signal and as a result, the data acquisition session for that channel.

For the load cell, the only anomaly that occurred during data acquisition session was the result of broken segments from the maxilla (be it cadaver or plastic) falling onto the head fixture and changing the load case for which the load cell was zeroed with thus giving a non-zero signal. This non-zero-signal occurred post-impact and had no influence on the pre-impact and impact data acquisition.

6.3 Modeling of the Anatomy and FE Simulation of Impact

6.3.1 Surface Modeling

Perhaps the largest challenge in the development of an FE model of a biological structure is the creation of a geometrically accurate representation of the existing anatomy. This challenge is further exemplified by the limitations imposed on the modeler due to the available software(s) and the available computing capacity. The raw data used in the creation of the model comes in the form of CT images of the anatomy in question. For the case of the maxilla, the available CT images contained blurring and artifacts that distorted the resulting image. The region of highest distortion was the anterior portion of the maxillary incisors, which is precisely the region of interest for this study. Using the thresholding and edge detection algorithms available in 3D Doctor™, along with a large amount of manual manipulation of NURBS curves from the operator, the general shape of the four anterior maxillary incisors was outlined and outputted to Rhino 3D™ modeling software. Here, with the assistance of plastic anatomical models and the guidance of Dr. David Kenny, the boundary contours for the four anterior maxillary incisors were manipulated until the desired shape and structure of four individual teeth was finally constructed. From here, the creation of four sided, isoparametric surfaces required the subdivision of the boundary contours into very small regions where the high curvature present in between the individual teeth could be captured. The assembled network of patches required over three hundred individual third-order NURBS surfaces in order to

accurately model the geometry. This process was extremely labour intensive and presented the single most time consuming portion of the entire project.

Sharp discontinuities such as the edges and points along the incisal edges of the teeth had to be avoided since this would lead to stress concentrations within the FE model, which would drastically alter the results of the impact simulations. Previous models of the craniofacial complex did not include dentition and therefore did not encounter this stress concentration issue (Krabbel & Appel, 1995; Horgan & Gilchrist, 2003; Voo *et al.*, 1996). In order to alleviate the stress concentrations present at regions of sharp discontinuity, edges and points were filleted and smoothed using Laplace smoothing algorithms with volumetric preservation. These algorithms remove local discontinuities without a change in the object's overall volume. Thus, the discontinuity is removed while maintaining the overall geometric integrity of the model. Statistically speaking, it is very difficult to quantify the overall error between a modeled jaw and the actual geometry (Korioth, 1992). The use of MSC.Patran's geometry fit error function served as a means of measuring the FE model's deviance from the actual anatomy thus confirming the model's ability to conform to the maxilla's anatomy.

6.3.2 Material Properties of Maxilla and Baseball

Currently, most available data regarding the anisotropic material properties of bone stems from the investigations done on long bones such as the femur and the tibia (Currey, 2002). The directionality of the material properties measured reveals distinct axes of strength along the long axis of the bone. This result agrees with intuition since it is this direction of the bone, which receives the main trajectory of the load. Unfortunately, this database does not extend well into bones of irregular structure, namely the maxilla. The curved geometry of the maxilla creates a very enigmatic set of material properties whose directionality has not yet been determined.

FE studies completed on the mandible have been able to use the existing data set by assuming that the U-shape of the mandible is simply a long bone bent around into a curve (Hart *et al.* 1992). This assumption preserves the established directionality of the long bone while applying it to a different geometry. This assumption once again, cannot be extended to that of the maxilla largely because there is no immediate intuitive directionality. There is no evidence that one particular direction within the structure should have any increase in strength as opposed to another. Rather than begin applying assumptions, which if not based on any empirical evidence would be the result of mere speculation, it was decided that the application of isotropic material properties would be a sufficient method of modeling the maxilla. Statistical analyses on the effect of variations within the

given set of material properties of a mandibular FE model have shown that changes of up to 50% in magnitude would be necessary in order to induce a significant change in the resulting strain pattern of the model (Korioth, 1992). The lack of distinct axes as a result of the geometry of the maxilla (resembling an aggregate of tissues rather than a beam or a column) suggests that such large variances in strength would not occur and therefore an isotropic set of material properties would suffice in capturing the mechanical behaviour of the system.

Current FE models of the entire skull (such as those by Krabbel & Appel, 1995; Horgan & Gilchrist, 2003; Voo *et al.*, 1996; Sujith, 2000) also employ isotropic material properties. Ideally, it would have been advantageous to measure the properties of the maxilla directly for this study, however, data from a concurrent study involving the mechanical testing of bone samples taken directly from the maxilla was unavailable at this time and therefore not employed.

The implementation of a failure criterion within the FE simulation is intimately related to the use of isotropic material properties. In order to implement even the simplest of failure criteria within MSC.Patran, the operator requires specific knowledge regarding the yield strengths along precise directions. Once again, the information required to fully accommodate the criteria necessary to implement a failure mode is simply not available for the maxilla (or any other oddly shaped bone) at this time. In order to preserve the integrity of the research it was decided to avoid simple guesswork and rely on the assumption that fracture of the maxilla occurred at the peak strain of the system. This peak strain could be easily found in both the experimental data and the FE simulation, which would employ linearly elastic, isotropic material properties.

Initially, the aim of the FE simulation was to incorporate both a deformable maxilla and a deformable baseball. This involved the implementation of isotropic material properties found in the literature. Recent investigations by Hendee *et al.*, 2000 and Smith *et al.*, 2000 have shown the mechanical behaviour of traditional baseballs both in static and dynamic situations. Using the materials properties found in their experimentation and modeling, a deformable model of the ball was created for the transient dynamic impact simulations. The results of the simulations showed significant deformation within the baseball itself. This deformation was an order of magnitude greater than the deformation induced within the maxilla itself. When examining the high-speed footage of the impact experiment on a frame-by-frame basis and taking relative measurements of both the ball and the maxilla deformations, it was found that the FE simulation was producing erroneous results. The deformation of the baseball throughout the impact profile was negligible compared to that of the maxilla. This prompted the investigation into the mechanical properties of the baseball. This

investigation was carried out by conducting some preliminary experiments involving the dropping of the ball onto a rigid surface and using the subsequent rebound to calculate the coefficient of restitution (*cor*) of the ball. The result of these preliminary investigations was in good agreement with the literature values but failed to provide any useful insight regarding the extreme deformations observed in the transient dynamic simulations. Therefore, it was decided that the application of the only empirical evidence available (in this case the results of the frame-by-frame examination of the high-speed footage) would be the most logical approach. As a result, further investigations into modeling the deformable nature of the ball were abandoned in favour of implementing a rigid ball model. This is arguably the soundest decision possible based on the available empirical data.

6.3.3 Validation of the FE Model Through Experimentation

Few FE studies of the biomechanics of the craniofacial complex have been validated through rigorous experimentation. In general, a comparison to literature or clinical values has served as the validation criteria. Studies by Iseri *et al.* (1998) involving the FE analysis of rapid maxillary expansion (RME) found continued difficulty in creating an experimental procedure which is capable of measuring the complex forces as moments which are the result of orthodontic treatment such as RME (Iseri *et al.*, 1998). The application of electrical resistance strain gauging or possibly holography is not sufficient to correctly depict, in real time, the detailed changes within the biological structures. This difficulty commonly arises when attempting to model the complex behaviour of biological systems and therefore many studies have not included the validation of their analysis through experimentation.

Studies by Verrue *et al.* (2001) and Anderson *et al.* (1999) have included rigorous experimentation as a means of validating their models. In studying the initial displacements as a result of orthodontic loads, Verrue *et al.* (2001) created a FE model of a canine skull and validated its response to orthodontic loads by carrying out an *in vivo* experiment with the same dog for which the model was based upon. Displacements induced in the dog's skull as a result of the experimentation were measured using laser speckle interferometry and subsequently used as a comparison to the values generated by their FE model (Verrue *et al.*, 2001). Similarly, Anderson *et al.* (1999) conducted an FE analysis of a sheep skull in order to investigate brain injury as a result of impact. An *in vivo* experiment was conducted on a sheep using an array of accelerometers and dynamic intracranial pressure transducers that were designed to measure the acceleration profile and the changes in

intracranial pressure during the impact (Anderson *et al.* 1999). The results of the experiment were then used to validate the outputted values from their preliminary FE model of sheep skull.

Although both the aforementioned studies are examples of validated FE investigations of the biomechanics of the craniofacial complex neither example provides a solid means of comparison for the present study. The experimental techniques employed relied on transducers and interferometry as their means of experimental measurement rather than the use of strain gauges. The application of these specific instruments to the investigation of the anterior portion of the human maxilla is simply not possible due to very strict space limitations. The use of accelerometers would not be as revealing as the use of strain gauges since it is the strain rate dependency of bone that has often been suggested as a dominating factor in its mechanical behaviour. By examining the impact profile captured by the strain gauges and conducting some routine calculations, an investigator could easily reveal data relating to the strain experienced by the specimen with respect to time. Employing accelerometers would not offer such a luxury. On the other hand, these studies offer specific means of comparing measured values with calculated ones. Statistically, the sample size of the present study's data is simply too small to perform any insightful statistical analyses. As a result, the only comparative means available to validate the FE model was the measurement of the relative differences in principal 2D strain on the chosen physiological landmarks.

Furthermore, the experimentation conducted by Anderson *et al.* (1999) and Verrue *et al.* (2001) was carried out *in vivo* and on animals. For obvious ethical reasons, the extension of their experimental protocol (particularly that employed by Anderson *et al.* (1999)) is simply not possible in the present study involving the analysis of human maxillæ. Therefore it is relatively safe to conclude that an *in vitro* study into the mechanical behaviour of the maxilla in response to an impact is perhaps the most robust means of validating a FE model at this time.

6.3.4 FE Models of the Human Maxilla and the Craniofacial Complex

Over the past two decades, several FE models of the craniofacial complex, both in 2D and 3D, have been developed and created for the purpose of understanding its mechanical behaviour (Iseri *et al.*, 1998; Gilchrist, 2003; Tanne *et al.*, 1988; Meyer *et al.*, 2001; Verrue *et al.*, 2001; Voo *et al.*, 1996; Krabbel & Appel, 1995; Anderson *et al.*, 1999). In general, these models focused on creating a simplified geometry of the entire craniofacial complex and the skull without placing much consideration on the dentition. In fact, many of these models drastically oversimplified the geometry of the maxilla by replacing its contours with simple linear segments. This ultimately places a limit on

the quality of the numerical approximation of the mechanical behaviour of the system. Also the majority of these models have employed shell elements to populate their meshes as opposed to solid continuum elements. The application of shell elements may be valid in regions of thin bone, such as the filamentary structures of the nasal bones where planar sections remain planar. However, volumes of solid bone occupy the bulk of the maxillofacial structure and therefore the application of shell theory, in particular its response to transverse shear loads, may not produce sufficiently accurate results (ABAQUS/Standard User's Manual Vol. 2, 2002). In order to support high shear loads, which are likely to occur as the result of an impact scenario, it is important to employ an element that is capable of absorbing that shear strain without requiring a very high mesh density in the process. Individual solid continuum elements are much more efficient at responding to shear loads than shell elements and therefore their application in the modeling of the maxilla seems wholly logical (ABAQUS/Standard User's Manual Vol. 2, 2002).

Chapter 7

7 CONCLUDING STATEMENTS AND SUMMARY

The recreation of blunt oral-facial trauma, through the use of experimentation on cadavers and a subsequent FE analysis of the human maxilla, is feasible. The collection of instrumentation available to the researcher at this time allows for the accurate acquisition of force, moment and strain data, without a significant influence on the system itself. Electrical resistance strain gauging has proved to be a reliable and valid technique for accurate strain measurement, in a variety of applications. These instruments, combined with a rigorous experimental protocol, have been shown to be fully capable at reducing a chaotic form of trauma into a well-controlled experiment yielding the same results. Much care must be exercised in order to ensure that ethical behaviour (and the preservation of a correct scientific method) is fully maintained throughout the experimental process, without risking injury or insult to anyone involved.

Using advanced imaging techniques, such as CT scanning or MRI, enables the researcher to capture the complex geometries of biological structures in a non-destructive and highly accurate way. The imaging of surface features, as well as the internal architecture, facilitates the creation of finely detailed solid models composed of simple mathematical functions. While the actual creation of the solid model is a somewhat labour intensive exercise (demanding a high level of skill from the operator) the resulting model exhibits a geometry that closely resembles the delicate curves and contours of the actual anatomy.

Transforming a geometrically accurate, solid model into a robust, mathematical one is not without its own set of unique challenges. Adaptive meshing and convergence testing are both techniques that allow the researcher to shape and manipulate the model in order to maximize accuracy and efficiency while minimizing computational time. Only a careful balance of these constraints, with a mindful application of appropriate boundary conditions and published material properties, will

result in a model that effectively simulates a complex dynamic event. Equally as important is the correct validation of the simulation. Validation can only be truly accomplished through a direct comparison with experimentally derived values. Thus, the intimate connection between experiment and model is revealed. The quality of one will invariably affect the measure of quality within the other and a substantial amount of thought is required to effectively marry these vastly different worlds in a harmonious manner.

In general, the patterns of strain revealed by both the experimentation and the subsequent FE model closely resemble clinical dental trauma. Highly localized trauma, centralized at the maxillary incisors, is in fact the commonest of all dental injuries and its presence within the investigation was no exception. The startling results discovered by closely examining the post-impact cadaver heads shows the severity of blunt oral-facial trauma, which re-enforces the motivation behind this investigation.

Ultimately, the conclusion that can be drawn from this study is that the FE method seems to be a robust and powerful method for modeling the biomechanics of the human maxilla in response to impact. The strong correlations with the experimental values confirm this notion and illustrate that the FE method can possibly be employed as a predictive tool for future investigations within this application. This extrapolation of the model is significant since several limitations are imposed concerning the extent of the experimentation. Ethics aside, it is unlikely that an experimental protocol will be established that is capable of revealing a complete collection of insights into the biomechanics of the maxilla simply because many of its most poignant features remain inaccessible to instrumentation. This is where the predictive ability of the FE method becomes invaluable and aside from the limitations of the computational resources available, it would seem that its application is bound only by the imagination of its user.

Chapter 8

8 FUTURE CONSIDERATIONS

It was well known at the onset of this investigation that boundaries would have to be imposed regarding the scope and detail of the project. This was due to the facilities available and the anticipated timeline for the completion of the current portion. The following paragraphs aim to illustrate the possible future considerations, which if applied, could lead to an overall improvement of the current FE model and the execution of further experimentation.

Firstly, the segregation of tissues varying from periodontal ligament to trabecular bone and dentin is crucial to understanding the more subtle behaviours of the maxillofacial system. Internal structures such as the root and pulp can be modeled and implemented into the existing model without compromising its performance if proper care be exercised. A fundamental understanding of the boundary conditions existing between the various tissues, particularly the interface between the periodontal ligament and the socket, is vital in order to ensure the preservation of the model's mathematical integrity. Knowledge of such boundary conditions is limited at this time and will require much effort to reveal their true nature.

The implementation of different materials and the assignment of varying material properties are also very important since the amount and nature of the deformations associated with a dynamic load will depend almost exclusively on their values. This is an important consideration since its application does not involve the mere implementation of different literature values. A complete experimental investigation is required in order to gain knowledge into the currently unknown anisotropic nature of bone tissue in the maxillofacial region. This could be accomplished by conducting linear and/or bilinear mechanical testing on maxillary bone samples along mutually orthogonal directions. This type of testing has been implemented in the mechanical testing of long

bones and if scaled down to the samples sizes available from maxillary tissues could be used to provide valuable insights. In the event the sample sizes available prove to be too small (since the maximum thickness of bone tissue from the maxillofacial complex is normally below 10mm) the use of acoustic, ultrasonic techniques conducted on intact maxillæ in an attempt to measure their mechanical properties may be the method of choice.

Similarly, the implementation of failure modes within the FE model would also prove to be advantageous. Once again though, since the literature is largely devoid of any rigorous experimentation of this nature, it is up to the experimenter to implement a protocol. The use of a slowly applied, linearly actuated load onto an intact maxilla until failure is induced may reveal the possible yield strengths of specific directions within the structure. A properly restrained specimen, in which no other portions of the sample are allowed to deform, is absolutely necessary in order to isolate the maxilla's mechanical behaviour in response to a well known load. The design challenges required to put into effect such an experiment are plentiful and difficult but certainly not unfeasible.

The inclusion of soft tissues such as the lips, tongue and periodontal tissues would also improve both the functionality and the aesthetics of the model. Several constitutive models employing viscoelastic or hyperelastic behaviour have been developed in the modeling of soft tissue (Ruan, 1994; Kleiven & Von Holst, 2002; Willinger *et al.*, 1995;). The damping effect induced by the inclusion of these tissues will most definitely affect both the strain patterns and magnitudes within the FE model. Most important of these soft tissues is the periodontal ligament. Unfortunately, at this time, the body of work regarding the investigation of the mechanical properties of the periodontal ligament lacks any strong inter-study correlation and therefore much is still left up to speculation. This, more often than not, comes at the expense of the model's integrity and therefore further investigations into the biomechanical properties of the periodontal ligament is desired.

Although difficult to implement, the creation of a full head model with a neck and associated musculature may ultimately prove to be the closest simulation to the actual events that occur during this type of trauma. The recoil effect caused by a head snapping backward as a result of a blunt impact may have significant effects on the observed trauma. In addition, this recoil may also be the cause of secondary spinal injuries, which occur simultaneously with the maxillofacial trauma.

Finally, as a predictive tool, a FE model of the human maxilla could be employed as a surrogate mouth in the testing of the next generation of mouth guards. A completed design could be fitted onto the FE model and a comparison of the peak strains as a result of a transient dynamic simulation could quantitatively evaluate the effectiveness of one design over another. If the aim of a

particular mouth guard design is to reduce the peak strains and maximum strain rate induced in a frontal impact to the maxillary incisors, then the existing FE model is capable of carrying out those tasks in a predictive and evaluative manner. As a design progresses in its evolution and features like user comfort, air passageway obstruction and aesthetics become more dominant than sheer function alone, innovative changes to the model would be required. Should these prescribed results come into effect, the role of dentist and engineer alike would then shift from a reactive, treatment nature to a more proactive and preventive one. Investigative work such as that presented within the body of this thesis ultimately serves as a stepping-stone towards the kind of research that prevents the tragedy of blunt oral-facial trauma. The author personally finds this notion extremely gratifying.

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Appendix A:

MATLAB Source Code

The following appendix is the source code employed for the data acquisition system used in all impact experiments. The data acquisition system software was divided into functions designed to complete specific tasks. These functions were then called upon by the higher order functions 'ready', 'fire' and 'cleanup' and passed variables by the experimenter prior to the data acquisition session.

Function for initializing the data acquisition boards for the strain gauges...

```
function[]=init_strain3(rate,test_time,strain_file)

%-----%
% This program will initialize a single data acquisition hardware board (aiSTRAIN)
% and setup its associated parameters.
% Note: This is currently setup for PCMCIA (cbi) boards.
%-----%

% Declaring all global variables.
global aiSTRAIN Index Overwrite

%-----%
% Setting up the strain gauge board.
%-----%

% Initializing the strain gauge board.
aiSTRAIN = analoginput('mcc',1);

% Specifying input type.
aiSTRAIN.InputType = 'SingleEnded';

% Adding channels to strain gauge board.
addchannel (aiSTRAIN,0:8);

% Giving each channel a unique name.
set(aiSTRAIN.Channel(1),'ChannelName','Strain_Gauge_1_Direction_1');
set(aiSTRAIN.Channel(2),'ChannelName','Strain_Gauge_1_Direction_2');
set(aiSTRAIN.Channel(3),'ChannelName','Strain_Gauge_1_Direction_3');
set(aiSTRAIN.Channel(4),'ChannelName','Strain_Gauge_2_Direction_1');
set(aiSTRAIN.Channel(5),'ChannelName','Strain_Gauge_2_Direction_2');
set(aiSTRAIN.Channel(6),'ChannelName','Strain_Gauge_2_Direction_3');
set(aiSTRAIN.Channel(7),'ChannelName','Strain_Gauge_3_Direction_1');
set(aiSTRAIN.Channel(8),'ChannelName','Strain_Gauge_3_Direction_2');
set(aiSTRAIN.Channel(9),'ChannelName','Strain_Gauge_3_Direction_3');
%set(aiSTRAIN.Channel(10),'ChannelName','Strain_Gauge_4_Direction_1');
%set(aiSTRAIN.Channel(11),'ChannelName','Strain_Gauge_4_Direction_2');
%set(aiSTRAIN.Channel(12),'ChannelName','Strain_Gauge_4_Direction_3');
%set(aiSTRAIN.Channel(13),'ChannelName','Strain_Gauge_5_Direction_1');
%set(aiSTRAIN.Channel(14),'ChannelName','Strain_Gauge_5_Direction_2');
%set(aiSTRAIN.Channel(15),'ChannelName','Strain_Gauge_5_Direction_3');

% Specifying how many seconds of data acquisition.
duration = test_time;

% Specifying sample rate in Hz.
set(aiSTRAIN,'SampleRate',rate);

% Specifying input ranges in volts.
set(aiSTRAIN.Channel(1:9),'InputRange', [-10,10]);

% Specifying sensor ranges in volts.
set(aiSTRAIN.Channel(1:9),'SensorRange', [-10,10]);

% Specifying units range in volts.
set(aiSTRAIN.Channel(1:9),'UnitsRange', [-10,10]);

% Getting the actual sample rate.
StrainActualRate = get(aiSTRAIN,'SampleRate');

% Setting the duration of the experiment in terms of samples taken per trigger.
set(aiSTRAIN,'SamplesPerTrigger', duration*StrainActualRate);
```

```

%-----%
% Setting up the trigger mode and all required properties
% to successfully log the data to workspace and file.
%-----%

% Setting the trigger type. Choose either immediate or manual.
set(aiSTRAIN, 'TriggerType', 'Manual');

% Setting the filename and filetype (.daq) for which the data will be recorded to.
set (aiSTRAIN, 'LogFileName', [strain_file, '.daq']);

% Setting the mode for which the data will be saved, in this case to the workspace
% as well as to the filename previously indicated.
set (aiSTRAIN, 'LoggingMode', 'Disk&Memory');

% This sets whether or not the log file is overwritten, or if multiple log files
% are written with an indexed name based on the value of LogFileName. Should we
% want to overwrite, simply replace 'Index' with 'Overwrite'.
set (aiSTRAIN, 'LogToDiskMode', 'Overwrite');

% Starting the hardware....
start(aiSTRAIN)

```

Function for initializing the data acquisition boards for the load cell...

```

function[]=init_load3(rate,test_time,load_file)
%-----%
% This program will initialize a single data acquisition hardware board (aiLOAD)
% and setup its associated parameters.
% Note: This is currently setup for PCMCIA (cbi) boards.
%-----%

% Declaring all global variables.
global aiLOAD Index Overwrite

%-----%
% Setting up the load cell board.
%-----%

% Initializing the load cell board.
aiLOAD = analoginput('mcc',0);

% Specifying input type.
aiLOAD.InputType = 'SingleEnded';

% Adding channels to load cell board.
addchannel (aiLOAD,0:5);

% Giving each channel a unique name.
set(aiLOAD.Channel(1), 'ChannelName', 'Load_cell_Fx');
set(aiLOAD.Channel(2), 'ChannelName', 'Load_cell_Fy');
set(aiLOAD.Channel(3), 'ChannelName', 'Load_cell_Fz');
set(aiLOAD.Channel(4), 'ChannelName', 'Load_cell_Mx');
set(aiLOAD.Channel(5), 'ChannelName', 'Load_cell_My');
set(aiLOAD.Channel(6), 'ChannelName', 'Load_cell_Mz');

% Specifying how many seconds of data acquisition.
duration = test_time;

% Specifying sample rate in Hz.
set(aiLOAD, 'SampleRate', rate);

% Specifying input ranges in volts.
set(aiLOAD.Channel(1:6), 'InputRange', [-10,10]);

```

```

% Specifying sensor ranges in volts.
set(aiLOAD.Channel(1:6), 'SensorRange', [-10,10]);

% Specifying units range in volts.
set(aiLOAD.Channel(1:6), 'UnitsRange', [-10,10]);

% Getting the actual sample rate.
LoadActualRate = get(aiLOAD, 'SampleRate');

% Setting the duration of the experiment in terms of samples taken per trigger.
set(aiLOAD, 'SamplesPerTrigger', duration*LoadActualRate);

%-----%
% Setting up the trigger mode and all required properties
% to successfully log the data to workspace and file.
%-----%

% Setting the trigger type. Choose either immediate or manual.
set(aiLOAD, 'TriggerType', 'Manual');

% Setting the filename and filetype (.daq) for which the data will be recorded to.
set (aiLOAD, 'LogFileName', [load_file, '.daq']);

% Setting the mode for which the data will be saved, in this case to the workspace
% as well as to the filename previously indicated.
set (aiLOAD, 'LoggingMode', 'Disk&Memory');

% This sets whether or not the log file is overwritten, or if multiple log files
% are written with an indexed name based on the value of LogFileName. Should we
% want to overwrite, simply replace 'Index' with 'Overwrite'.
set (aiLOAD, 'LogToDiskMode', 'Overwrite');

% Starting the hardware....
start(aiLOAD)

```

Function for readying the data acquisition software...

```

%-----%
% Function to setup a data acquisition session for two boards,
% and all their variables.
%-----%
% Author: Peter Krimbalis.
% Date: August 16, 2002.
%-----%

function []=ready(strain_file,load_file)

% Declaring all global variables needed for the entire acquisition session
global aiSTRAIN aiLOAD Index Overwrite
global strain_data load_data strain_time load_time strain_abstime
global load_abstime strain_events load_events

% Initializing the boards, setting sample rate and duration of acquisition
init_strain3(6250,5,strain_file)
init_load3(6250,5,load_file)

```

Function for triggering the strain gauge data acquisition board...

```

%-----%
% Program to Start Data Acquisition via a Manual Trigger
%-----%

function trigger_strain()

```

```

% Declaring all global variables
global aiSTRAIN

% Firing the manual trigger....
trigger(aiSTRAIN)

%-----%
% Data should now be logging to workspace and file. The duration of this logging,
% and any other similar properties depend entirely on the parameters established
% in the m-file called 'initialize_boards.m'
%-----%

% Stopping the hardware....
%stop(aiSTRAIN)

```

Function for triggering the load cell data acquisition board...

```

%-----%
% Program to Start Data Acquisition via a Manual Trigger
%-----%

function trigger_load()

% Declaring all global variables
global aiLOAD

% Firing the manual trigger....
trigger(aiLOAD)

%-----%
% Data should now be logging to workspace and file. The duration of this logging,
% and any other similar properties depend entirely on the parameters established
% in the m-file called 'initialize_boards.m'
%-----%

```

Function for retrieving acquired strain gauge data from the MATLAB acquisition engine...

```

%-----%
% Program to retrieve data for one board from data acquisition engine.
%-----%

function retrieve_strain(samples)

% Declaring all global variables
global aiSTRAIN strain_data strain_time strain_abstime strain_events

% Retrieving the data from the data acquisition engine
[strain_data, strain_time, strain_abstime, strain_events]=getdata(aiSTRAIN, samples);

```

Function for retrieving acquired load cell data from the MATLAB acquisition engine...

```
%-----%
% Program to retrieve data for one board from data acquisition engine.      %
%-----%

function retrieve_load(samples)

% Declaring all global variables
global aiLOAD load_data load_time load_abstime load_events

% Stopping the hardware....
%stop(aiLOAD)

% Retrieving the data from the data acquisition engine
[load_data,load_time,load_abstime,load_events]=getdata(aiLOAD,samples);
```

Function for plotting acquired strain gauge data, post impact...

```
%-----%
% PLOT_STRAIN3: plotting of data + titling.                                %
% Input:  data [frame,channel]                                             %
% Output:                                                                    %
%-----%
% Author: Peter Krimbalis                                                  %
% Date: May 6, 2003.                                                       %
%-----%

function []=plot_strain3(data)

% Declaring all the global variables.
global aiSTRAIN strain_data strain_time strain_abstime strain_events

% Plotting the data from gauge 1.
figure,
colordef white
subplot(2,2,1), plot(strain_data(:,1))
grid on
title('Gauge 1, Direction 1')
subplot(2,2,2), plot(strain_data(:,2))
grid on
title('Gauge 1, Direction 2')
xlabel('Samples')
ylabel('Voltage')
subplot(2,2,3), plot(strain_data(:,3))
grid on
title('Gauge 1, Direction 3')

% Plotting the data from gauge 2.
figure,
colordef white
subplot(2,2,1), plot(strain_data(:,4))
grid on
title('Gauge 2, Direction 1')
subplot(2,2,2), plot(strain_data(:,5))
grid on
title('Gauge 2, Direction 2')
xlabel('Samples')
ylabel('Voltage')
subplot(2,2,3), plot(strain_data(:,6))
grid on
title('Gauge 2, Direction 3')
```

```

% Plotting the data from gauge 3.
figure,
colordef white
subplot(2,2,1), plot(strain_data(:,7))
grid on
title('Gauge 3, Direction 1')
subplot(2,2,2), plot(strain_data(:,8))
grid on
title('Gauge 3, Direction 2')
xlabel('Samples')
ylabel('Voltage')
subplot(2,2,3), plot(strain_data(:,9))
grid on
title('Gauge 3, Direction 3')

% Plotting the data from gauge 4.
figure,
colordef white
subplot(2,2,1), plot(strain_data(:,10))
grid on
title('Gauge 4, Direction 1')
subplot(2,2,2), plot(strain_data(:,11))
grid on
title('Gauge 4, Direction 2')
xlabel('Samples')
ylabel('Voltage')
subplot(2,2,3), plot(strain_data(:,12))
grid on
title('Gauge 4, Direction 3')

% Plotting the data from gauge 5.
figure,
colordef white
subplot(2,2,1), plot(strain_data(:,13))
grid on
title('Gauge 5, Direction 1')
subplot(2,2,2), plot(strain_data(:,14))
grid on
title('Gauge 5, Direction 2')
xlabel('Samples')
ylabel('Voltage')
subplot(2,2,3), plot(strain_data(:,15))
grid on
title('Gauge 5, Direction 3')

% Finding the maximum and minimum value for each channel.
for i=1:9
    strain_max_voltage(i)=max(strain_data(:,i));
end

for j=1:9
    strain_min_voltage(j)=min(strain_data(:,j));
end

% Printing the results on the command window.
strain_max_voltage
strain_min_voltage

%-----%
% All 15 channels should now be plotted on 5 multiplot figures.
%-----%

```

Function for plotting acquired load cell data, post-impact...

```

%-----%
% PLOT_LOAD3: plotting of data + titling.                                     %
% Input: data [frame,channel]                                                %
% Output:                                                                    %
%-----%
% Author: Peter Krimbalis                                                    %
% Date: May 6, 2003                                                         %
%-----%

function[]=plot_load3(data)

% Declaring all the global variables.
global aiLOAD load_data load_time load_abstime load_events

% Plotting the force data.
figure,
colordef white
subplot(2,2,1), plot(load_data(:,1))
grid on
title('Force in X Direction')
subplot(2,2,2), plot(load_data(:,2))
grid on
title('Force in Y Direction')
xlabel('Samples')
ylabel('Voltage')
subplot(2,2,3), plot(load_data(:,3))
grid on
title('Force in Z Direction')

% Plotting the moment data.
figure,
colordef white
subplot(2,2,1), plot(load_data(:,4))
grid on
title('Moment About X Axis')
subplot(2,2,2), plot(load_data(:,5))
grid on
title('Moment About Y Axis')
xlabel('Samples')
ylabel('Voltage')
subplot(2,2,3), plot(load_data(:,6))
grid on
title('Moment About Z Axis')

% Finding the maximum and minimum value for each channel.
for i=1:6
    load_max_voltage(i)=max(load_data(:,i));
end

for j=1:6
    load_min_voltage(j)=min(load_data(:,j));
end

% Printing the results on the command window.
load_max_voltage
load_min_voltage

%-----%

```



```
% All 6 channels should now be plotted on 2 multiplot figures.
%-----%
```

Function to actually initiate a data acquisition session...

```
%-----%
% Function to begin acquiring data for two boards, and plot the results.
%-----%
% Author: Peter Krimbalis.
% Date: August 16, 2002.
%-----%

% Declaring all global variables
global aiSTRAIN aiLOAD filename Index Overwrite
global strain_data load_data strain_time load_time strain_abstime
global load_abstime strain_events load_events

% Pulling the trigger....
trigger_strain
trigger_load

% Extracting data from the engine...
retrieve_load(31250)
retrieve_strain(31250)

% Plotting the results on separate figures...
plot_load3(load_data)
plot_strain3(strain_data)
```

Function to purge all used variables and end the data acquisition session...

```
%-----%
% Function to end a data acquisition session for two boards and reset them.
%-----%
% Author: Peter Krimbalis.
% Date: May 7, 2003.
%-----%

% Declaring all global variables needed for the entire acquisition session
global aiSTRAIN aiLOAD strain_file load_file Index Overwrite
global strain_data load_data strain_time load_time strain_abstime
global load_abstime strain_events load_events

% Resetting the hardware....
stop (aiSTRAIN)
stop (aiLOAD)
```

Function for calculating the actual load experienced by the load cell...

```
%-----%
% Function to calculate the actual load experienced by the load cell
% Input:
% amp = amplification values[6x1]
% data = array of amplified data values [samples,signals]
% rate = sampling frequency
%-----%
% Author: Peter Krimbalis with assistance from Alan Moris
% Date: May 7, 2003.
%-----%

% Declaring the function.
```

```

function[actual_load]=calc_load(amp,data,rate)

% Establishing the calibration matrix as given by Bertec.
C = [451.7 37.3 -2.5 -2.5 1.5 -2.9; ...
     26.6 453.1 -1.0 -4.3 -0.5 0.1; ...
     11.7 10.6 378.2 -3.6 -4.2 -11.4; ...
     -1.1 -16.4 -0.2 5.6 -0.1 0.0; ...
     16.2 1.5 0.0 -0.1 5.6 -0.2; ...
     0.2 0.0 0.0 0.0 0.0 8.4];

% Setting the size according to the size of the data set.
[samples]=size(data,1);
[num_sigs]=size(data,2);

% Calculating the actual force on the load cell by first
% dividing out the amplification value and then
% multiplying by the calibration matrix.
for i=1:samples
    amp_signals(1:6,1) = data(i,1:6)';
    raw_signals = amp_signals./amp;
    % Calculates load value from C*raw_signals
    % transposes vector of actual loads into a row
    % for each sample of data [Fx Fy Fz Mx My Mz] [N.m].
    actual_load(i,1:6) = (C*raw_signals)';
end

% Plotting the results on separate graphs.
figure,
colordef white
plot(actual_load(:,1))
grid on
title('Force in X Direction')
xlabel('Samples')
ylabel('Force [N]')

figure,
colordef white
plot(actual_load(:,2))
grid on
title('Force in Y Direction')
xlabel('Samples')
ylabel('Force [N]')

figure,
colordef white
plot(actual_load(:,3))
grid on
title('Force in Z Direction')
xlabel('Samples')
ylabel('Force [N]')

figure,
colordef white
plot(actual_load(:,4))
grid on
title('Moment About X Axis')
xlabel('Samples')
ylabel('Moment [N.m]')

figure,
colordef white
plot(actual_load(:,5))
grid on
title('Moment About Y Axis')
xlabel('Samples')
ylabel('Moment [N.m]')

```

```
figure,  
colordef white  
plot(actual_load(:,6))  
grid on  
title('Moment About Z Axis')  
xlabel('Samples')  
ylabel('Moment [N.m]')
```

```
%-----%  
% The actual load experienced by the load cell should now be plotted on 6 %  
% separate graphs. %  
%-----%
```

Appendix B:**Ancillary Experimental Values and Modeling Tabulations**

The following appendix includes all the remaining strain gauge and load cell values from all experiments, which were not presented in the main body of the text. Also included are any other tabulated values and calculations made regarding the processing of the data and its use as validation for the FE model of the maxilla.

Cadaver 5164:

Cadaver 5164 Continued:

Cadaver 5164 Continued:

Cadaver 5180:

Cadaver 5180 Continued:

Cadaver 5180 Continued:

Cadaver 5186:

Cadaver 5186 Continued:

Cadaver 5186 Continued:

Cadaver 5212:

Cadaver 5212 Continued:

Cadaver 5212 Continued:

Cadaver 5247:

Cadaver 5247 Continued:

Cadaver 5247 Continued:

Location	Node Number	x co-ordinate	y co-ordinate	z co-ordinate
R. Canine Fossa	19560	-0.010625567	-0.018901501	-0.001730934
Medial Palate	12605	0.003362551	-0.016332578	-0.001751195
L. Canine Fossa	22396	0.012720513	-0.017576715	0.000869851

Convergence Tabulations:

Data Common to All Models						
Locations of Interest (from Geo)	Co-ord 1	Co-ord 2	Co-ord 3	Node #		
Point 1 (# 43)	0.002782075	-0.022523304	-0.008185478	1		
Point 2 (# 58)	0.002548365	-0.018070003	-0.008185478	2		
Point 3 (# 80)	0.020321423	-0.012209106	0.000877022	3		
Point 4 (# 83)	-0.016760815	-0.01406466	0.000868954	4		
Points of Fixation (All DOF)	Co-ord 1	Co-ord 2	Co-ord 3	Node #		
Point 1 (# 94)	0.004384787	-0.018056475	0.007595772	5		
Point 2 (# 25)	0.020415623	-0.010313219	0.007595772	6		
Point 3 (# 89)	-0.014955571	-0.013984037	0.007597724	7		
Point 4 (# 150)	-0.010955269	0.025353856	0.006033273	8		
Point 5 (#151)	0.009386198	0.025353856	0.006033273	9		
Load Data	Co-ord 1	Co-ord 2	Co-ord 3			
Point of Application (# 43)	0.002782075	-0.022523304	-0.008185478			
Point of Application (# 87)	-0.016041692	-0.014151781	0.005095772			
Point of Application (# 92)	0.02038784	-0.011216786	0.005095772			
Magnitude	0	100	0			
Material Props Used	Value	Element Type	10 node tetrahedral			
Young's Modulus	2.00E+10					
Poisson's Ratio	0.33					
Model Specific Data						
Model Name	Base Units	# Nodes	# Elements	Global Edge Length	Edge Length Factor	Actual Edge Length Minimum
converge1	metres	3215	1838	0.016279	0.4	0.0065116
converge2	metres	4135	2377	0.016279	0.3	0.0048837
converge3	metres	7389	4408	0.016279	0.2	0.0032558
converge4	metres	11524	7007	0.016279	0.15	0.00244185
converge5	metres	24616	15753	0.016279	0.125	0.002034875

Hart used an average stiffness for cortical bone of 15GPa and density of 2000 kg/m³

Futterling used an average stiffness for cortical bone of 14.7 and density of 0.0013 g

Kober used 11GPa and poisson's on 0.28

Hart used an average stiffness for cortical bone of 15GPa and density of 2000 kg/m³
 Furlerling used an average stiffness for cortical bone of 14.7 and density of 0.0013 g
 Kober used 11GPa and poisson's on 0.28

Model Name	Equivalent	Duplicates	Connectivity	Max Jacobian Ratio	Max Jacobian Zero
converge1	Yes	None	Yes	194	0.000000548
converge2	Yes	None	Yes	7.10E+01	2.93E-07
converge3	Yes	None	Yes	8.42E+00	1.61E-07
converge4	Yes	None	Yes	5.92E+00	1.58E-07
converge5	Yes	None	Yes	5.4	1.53E-07

Model Name	Geometry Fit Error	Max Stress	Node #	Min Stress	Node #
converge1	0.00145	3.63E+07	3002	29900	2530
converge2	1.38E-03	2.36E+07	3880	1.82E+04	3202
converge3	1.04E-03	4.34E+07	11	1.70E+04	2888
converge4	1.01E-03	8.22E+07	83	1.82E+04	7541
converge5	1.01E-03	8.23E+07	240	1.82E+04	7541

Model Name	Disp @ Point 1	% change	Disp @ Point 2	% change	Disp @ Point 3	% change
converge1	9.8016E-06	N/A	2.8073E-06	N/A	3.2158E-06	N/A
converge2	1.10E-05	12.68568397	3.8593E-06	37.47372921	3.2988E-06	2.581006281
converge3	1.27E-05	15.27387958	4.42E-06	14.46376286	4.0273E-06	22.08378804
converge4	1.31E-05	2.756833176	4.7515E-06	7.560837578	4.22E-06	4.792292603
converge5	1.33023E-05	1.67591531	4.90E-06	3.054824792	4.28E-06	1.492879653

Model Name	Max Displacement	Node #	% change	Total DOF's in Model	% change
converge1	0.0000141	2065	N/A	9645	N/A
converge2	1.71E-05	1	21.27659574	12405	28.61586314
converge3	1.98E-05	5264	15.78947368	22167	78.69407497
converge4	1.99E-05	1	0.505050505	34572	55.96156449
converge5	1.99E-05	1	0.113266332	73848	113.6063867