ROLE OF THE PERIODONTAL LIGAMENT IN OCCLUSAL LOAD TRANSFER: IMPLICATIONS FOR FINITE ELEMENT MODELS

By

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The periodontal ligament (PDL) is the medium of occlusal force transfer in the jaws and its mechanical properties likely have an effect on masticatory strain dissipation in the skull. The importance of incorporating the PDL in theoretical modeling is unexplored and the PDL is often ignored in finite element analyses of the primate mandible and cranium. Experimental data were collected to evaluate the effect of PDL behavior in four contexts: 1) to compare the strain response of a cranium to a load applied through an intact PDL versus a load applied through simulated direct tooth-bone contact, 2) to consider the effects of the PDL in strain dissipation local to and remote from the point of load application and 3) to establish the degree of rate dependant strain behavior observed in alveolar bone and 4) to evaluate the degree of hysteresis observed in alveolar bone in the two loading environments.

In vitro strain analysis was performed on a previously frozen, fresh *Sus domesticus* cranium. A vertical occlusal load at the left first deciduous molar was applied at varied rates and surface bone strain data were collected from rectangular rosette strain gages bonded to the cranium with the PDL intact and subsequently with a tooth-bone interface established using an adhesive with properties comparable to cancellous bone (cyanoacrylate). The viscoelastic properties and rate dependency of the PDL affected strain dissipation in the adjacent alveolar

bone. The cyanoacrylate interface produced markedly different strain magnitudes; however, we infer that effects of the PDL are primarily local to the load application site as remote gages did not register significant differences in strain between iterations.

CHAPTER 1
INTRODUCTION

Models of the masticatory system are important in interpreting the relationship between function and form of the dental apparatus. In generating these models there remains a level of uncertainty as to the degree of detail required in modeling the dentition and supporting structures. This study addresses the role of the periodontal ligament (PDL) in determining strain profiles of the alveolar region. Tendons, ligaments and bone are specialized to resist forces in specific directions and are developed and maintained under the loading histories to which they are exposed (Carter and Beaupre 2001). The PDL is the medium of occlusal force transfer in the jaws and its mechanical properties have an unknown effect on masticatory strain dissipation in the skull.

The functional adaptation of bone to its loading history is best understood by evaluating internal and surface distributions of stresses and strains created by applied loads (Carter and Beaupre 2001). In this study, in vitro strain gage techniques are used to analyze facial strain as a result of occlusal loading with and without an intact PDL in a porcine specimen. The PDL may act as a shock absorber for occlusal loads (Matsuo and Takahashi 2002), however, this behavior is not well-defined in the clinical literature. The PDL acts as a hydraulic damper serving to reduce the magnitude of loads transferred to the surrounding alveolar bone (Nabuto et al. 2003). Based on this behavior, it is anticipated that the strain magnitudes and distribution in the alveolus will be altered by removal of the PDL. Additionally, the PDL is a highly viscoelastic material (Toms et al. 2002a, 2002b). If the mechanical properties of the PDL do affect alveolar bone response to load, a reduction in viscoelastic behavior is expected after PDL removal. The goal of these experiments is to evaluate the role of the PDL in facial strain distribution. A more complete understanding of the interaction between alveolar bone and the PDL can inform accurate computer models of mastication mechanics and help establish the level of detail required in development and testing of these models.

Morphology of jaws and teeth vary among mammals. Dental and other masticatory characters of extant animals strongly correlate with dietary adaptations and their morphology is often extrapolated out to the fossil record. Primate evolution is characterized by many changes in the chewing apparatus, for example, strepsirhine tooth combs, the anthropoid fused mandibular symphysis, and the emergence of the chin in humans. To accurately appreciate the selective force acting on these features requires a structural and functional awareness of the components of the masticatory system. Part of this process involves a clear understanding of the load pattern resulting from jaw activity (Hiiemae 1984).

Bone morphology is influenced by its mechanical environment; accordingly mastication mechanics can provide insight into the evolutionary history of animals (O'Connor et al. 1982, Lanyon and Rubin 1984, Rubin and Lanyon 1985). Skull form is affected by the size and muscle attachment of the masticatory apparatus (Ringqvist 1973). Interaction of teeth and food combined with chewing force and structural resistance create unique loading patterns in a skull based on chewing characteristics (Herring 1993). Stress resulting from mastication is confined to the head, and is thought to affect the thickness and form of facial bones (Lavelle et al. 1977, Hiiemae and Crompton 1985).

To understand the mechanical consequence of the PDL in mastication the behavior of the bony components of the jaw need to be evaluated using stress and strain. Stress and strain are two central concepts in the study of the effect of mechanical forces on bone. Using these concepts we can understand morphological variation in masticatory structures. Stress is defined as force per unit area; units are generally MPa (mega Pascal, equivalent to N/mm^2). It is a quantitative measure of the intensity of internal forces. Strain is the measure of the local deformation at a point in a material, the fractional changes in dimension of a loaded body, and is dimensionless. It tells us by what proportion an object is being deformed. Stress and strain are vector quantities having both a magnitude and a direction (Carter and Beaupre 2001).

A comprehensive picture of the two and three dimensional strain state requires consideration of both normal and shear strains. Normal strains are the relative change in the length of the side of a model cube. Shear strain is the change in angle between two sides of the model cube that were initially perpendicular (Carter and Beaupre 2001). The components of strain are dependent on the coordinate system in use. A coordinate system can be found in which all shear strain components vanish, this defines the principal strain directions. The principal directions are important to establish considering that micro- and ultra-structural features like osteons, trabeculae and collagen fibers generally align with these strain orientations (Carter and Beaupre 2001).

Studies of strain are routinely done using a type of transducer; in the present experiments strain gages are used. A strain gage is an electrical resistance device used to measure surface strain over a finite area on a structure. When bonded to the surface the gage measures strain by stretching or contracting with the specimen to which it is bonded. An electrical circuit is used to measure the change in voltage which is then converted into a measure of strain.

Bone responds to stress and strain stimuli. In the chewing apparatus, different masticatory behaviors result in different levels and types of stress that influence the growth and development of bone (Frost 2004). Mechanical input affects bone morphology by stimulating or suppressing modeling and remodeling processes. Modeling generally affects the size and shape of bone whereas remodeling typically does not. Both processes involve osteoclasts and osteoblasts;

however, in modeling the two cell types work independently, in remodeling they are coupled (Martin et al. 1998). Modeling and remodeling are similar processes and both are most active during skeletal development. Once maturity is reached the rate of modeling versus remodeling declines. Strain rate, strain magnitude and strain distribution all interact to affect the rate and magnitude of bone remodeling (Lanyon 1984). Most experimental studies focus on remodeling.

The goal of this study is to clarify the role of the PDL in mastication, specifically in reference to its place in computer models of the face and jaws. The experimental data presented here serves to address multiple objectives: 1) to contrast the strain response of a cranium to a load applied through an intact PDL versus a load applied through simulated direct tooth-bone contact, 2) to evaluate the effects of the PDL local to and remote from the point of load application and 3) to ascertain the degree of rate dependent strain behavior observed in alveolar bone and 4) to assess the degree of hysteresis detected in alveolar bone in the two loading environments.

What follows is a discussion of alveolar bone morphology and mechanical properties as they relate to the PDL and mastication in Chapter 2. Chapter 3 addresses PDL development and mechanics and its various masticatory roles. Of particular importance to this project are the integrated roles of the PDL and alveolar bone in maintaining a functional jaw. Chapter 4 is a review of finite element modeling practices and includes an evaluation of studies relevant to the role of the PDL from the clinical and comparative perspectives. Together these chapters provide the foundation for the experimental data which serves to elucidate the mechanical role of the PDL in alveolar bone response and outline the level of detail to which these attributes should be addressed in future models.

CHAPTER 2
ALVEOLAR BONE

Before birth, gene expression in utero outlines initial bony anatomy, relationships and biological mechanisms that allow bone to adapt after birth (Frost 2004). The loading history of a bone is the record of forces put on the bone during a period of time. Loads on bone cause strains that generate signals specialized bone cells can detect and respond to. Mechanical studies of the skeleton rely on the assertion that the loading history of a bone has an effect on the shape of that skeletal element (Carter and Beaupre 2001). Bone's interpretation of load signals is mediated by genetically determined thresholds that control modeling and remodeling. If bone strains are lower than the threshold disuse remodeling may begin to reduce whole bone strength. If bone strains exceed the established threshold, remodeling may begin to strengthen bone (Frost 2004). Together genetics and the mechanical environment shape the strength of load bearing bones.

Morphology

Teeth are found in bony sockets in the alveolar processes of the maxilla and mandible. Several different components make up the functional alveolar unit. The tooth socket is lined by a thin lamella of bone. This region is the cribiform plate or alveolar bone proper and contains the inserting Sharpey's fibers of the PDL. A thicker outer layer of bone lines the lingual and labial surfaces of the alveolar process. This outer layer of bone is comprised of outer cortical plates and inner spongy bone; it directly surrounds the cribiform plate of alveolar bone. These outer layers are also referred to as basal or laminar bone (Bartold and Narayanan 1998). Schroeder (1992) defines the alveolar bone as bundle bone, a type of bone that contains several layers of bone in an orientation parallel to the tooth socket wall, which has Sharpy's fibers emerging from it at right angles.

Mechanical Properties

For modeling purposes the alveolar process is generally split into cancellous and cortical bone. Cancellous bone refers to the bundle or alveolar bone and has been assigned a Young's modulus between 272 and 489 MPa (Martin et al. 1998) and a Poisson's ratio of 0.31 (Jones et al. 2001). The bone surrounding bundle bone is cortical bone and usually assigned a modulus of ~13,800 N/mm^2 and a Poisson's ratio of 0.26 (Jones et al. 2001). McHlaney (1966) found that response of cortical bone is somewhat strain rate sensitive. Asundi and Kishen (2000) compared the strain distributions of tooth root and the supporting bone. They found higher strains in the cervical region and reduced strains in the apical region. This suggests that the PDL and surrounding bone are intricately linked in the distribution of masticatory forces.

Development

The alveolar bone is formed during root development and is derived from osetoblasts originating in the dental follicle. Its development is closely associated with the development of both the teeth and PDL. Alveolar bone proper develops independently of other components of the alveolar process. Primary and permanent teeth develop alveolar bone around their roots during development and eruption; succedaneous, or secondary, teeth develop alveolar bone later. Initially, deciduous and succedaneous teeth precursors are housed in the same osseous cavity (Moss-Salentjin and Hendrick-Klyvert 1990). As the deciduous teeth erupt alveolar bone surrounds the developing roots and separates the erupted deciduous tooth from the developing secondary tooth. As a result, succedaneous teeth become enclosed in a bony crypt after deciduous teeth erupt. During the eruption of a succedaneous tooth the walls of the bony crypt, the roots of the deciduous tooth and the alveolar bone housing the primary tooth are all resorbed. Only after the secondary tooth moves into place does new alveolar bone form around the newly erupting tooth (Bartold and Narayanan 1998). The alveolar bone of the maxilla and mandible begin to remodel upon emergence of the secondary teeth. The process of eruption of secondary teeth results in the complete deposition of new alveolar bone accompanied by major remodeling of the whole alveolar process.

Alveolar Remodeling

In addition to the remodeling that results from tooth eruption, the tooth commonly drifts mesially during the life of an individual. When a tooth moves, the bone its path is resorbed to make room for the advancing tooth. Bundle bone is broken down initially which results in the breakdown of PDL fibers inserting there. In some cases resorption may extend into lamellar bone. After enough resorption has occurred new bone fills in the space and reanchors PDL fibers and becomes new bundle bone. The side from which the tooth is moving undergoes bone deposition to fill in the space over which the tooth has moved. Since new bone is being deposited around PDL fiber it too becomes bundle bone. Continuous remodeling processes remove the deeper layers of bundle bone and replace it with lamellar bone. This occurs simultaneously around the perimeter to maintain a standard width around the tooth root. (Moss-Salentijn and Hendricks-Klyvert 1990). Orthodontic tooth movement takes advantage of the natural drift remodeling process.

Whenever a bolus, a mass of chewed food, is loaded between the teeth both the teeth and surrounding bone are deformed. Bone tissue remodels in response to these loads, but strain thresholds for deposition and resorption have not been reliably determined for the masticatory system (Biewener 1992). The same general findings of post-cranial bone remodeling stimuli are generally considered to apply to the jaws. However, the skull may behave somewhat differently from the post-crania in that recent studies have found that no local stimulus was required for skull bone deposition. In an armadillo case cranial bone thickened in response to general exercise, apparently independent of mechanical loading during mastication (Lieberman 1996).

However, there could be low level strains produced by impact vibrations and ground reaction forces of a sufficient frequency to produce cranial bone remodeling (Judex 2006). On the whole, the effects of loading on alveolar remodeling are even less clear than the effects of loading of bone in general.

It has been established, however, that the PDL plays an integral role in maintain alveolar bone. Work to date on the PDL suggests that it is the medium of force transfer and it directs the alveolar socket wall to remodel in response to applied forces. The PDL distributes applied force to alveolar bone and the frequency, direction and duration and size of the forces applied affects the extent and rate of bone remodeling. Most importantly forces applied to teeth lacking a PDL produce less and slower bone remodeling (Beertsen et al. 2000). This is supported by the relative failure of osseous implants to produce bone remodeling (Brunski 1999).

While it is clear that applied mechanical force produces bone remodeling in the alveolus, the exact nature of the interaction between ligament and bony tissue in force transfer and mechanisms that induce remodeling are unknown. When PDL cells are removed ankylosis sometimes occurs. The PDL space is invaded by bone tissue and creates a direct connection between the wall of the alveolar socket and the tooth (Beertsen et al. 2000). If PDL fibroblasts are inserted in an anklyosed area the PDL can repopulate the damaged area, but they must be preceded by cells that have the ability to resorb bone or cementum. The molecular mechanisms and signals of this process are unclear but eventually areas of bone can be replaced by a normally functioning and structured PDL. Additionally, mastication loads can increase the speed at which anklyosed areas are restored after insertion of PDL cells (Anderssen et al.1985, Wesselink and Beertsen 1994).

Osseointegrated prostheses provide an excellent test of bone remodeling behavior in the absence of PDL. Generally the implants function unlike the natural tooth-PDL-bone interface and often result in bone resorption. It is thought that the close association of bone and implant material results in the interface moving as a unit. This whole unit movement does not mimic the normal motion of tooth and bone relative to one another and may result in increased load transfer to all parts of the interface (Skalak 1983). Recent attempts to mimic the shock absorbing qualities of the PDL have found that inserting a pliable material with the implant can attenuate peak loads, reducing the amount of force transferred to alveolar bone during tooth loading (Carvalho et al. 2004), however, it is not clear what effect this has on remodeling. What is clear is the close relationship between the PDL and associated bone. To further understand the load histories represented by alveolar morphology it is critical to understand its mechanical relationship with the PDL.

CHAPTER 3
PERIODONTAL LIGAMENT

The periodontal ligament (PDL) functionally links the teeth to the alveolar bone. It provides support, protection and sensory input to the masticatory system. The PDL allows the teeth to change position under loading and maintains continuity between the two hard tissue components of the periodontal region, the alveolus and tooth cementum. In addition to its role as a support structure it also transmits blood vessels, supplies nourishment to the cementum, and maintains and remodels the soft and hard tissue of the periodontal region. The PDL is a unique tissue and removal of it can cause loss of function of teeth and resorption of surrounding alveolar bone. In mastication its role is to provide sensory feedback during the chewing cycle and to serve as a medium of force transfer between the teeth and alveolus.

Development of the PDL

The PDL is produced before dental eruption mainly from fibroblasts that originate in the dental follicle and begin to differentiate during root development (Ten Cate et al. 1971). During apical development the cells of the dental follicle differentiate into cementoblasts and form the cementum lining of the root surface (Grant and Bernik 1972).

Initially collagen fibers become embedded in the cementum and Sharpey's fibers are laid down in a coronal direction within the PDL region. The initial orientation is nearly parallel to the root surface. Fiber formation and deposition occurs from the developing cementoenamel junction (CEJ) to the apex of the tooth. Those fibers deposited apical to the CEJ form the PDL. After one-third of root formation fibers insert within the cementum matrix from the CEJ and continue coronally. This process closely follows the outline of the newly formed crown. No insertion of collagen fibers into the alveolar bone can be seen at this stage.

Loosely arranged fiber deposition and insertion continues along the developing root surface. Opposite this surface, fiber insertion also occurs along the lining of the bony socket wall. These fibers cross the ligament space in a similar fashion as the root side fibers. Both root and bone side fibers will ultimately come together in the middle of the ligament space to form the intermediate plexus. The orientation of these fibers is initially parallel to the root surface but as the teeth erupt their orientation dramatically changes (Grant and Bernik 1972) possibly as a result of the positional relationship of the erupting tooth to the adjacent teeth (Bartold and Narayanan 1998).

During eruption the dentogingival fibers align from the CEJ in the occlusal direction terminating in the connective tissue of the gingiva. The transseptal fibers extend over the alveolar crest in an oblique direction toward the surface of the adjacent developing tooth root. The fibers of the cervical-most one-third of the root surface run obliquely in the apico-occlusal direction from cementum to bone. Although there is still no direct connection from the root and bone fibers in the middle third of the root, they become more defined. Root formation is still occurring in the apical portion and accordingly fiber arrangement is poorly developed.

Upon full eruption and occlusal contact the ligament fibers assume their final configuration. The dentogingival, transseptal and alveolar crest fibers originate at the CEJ. Within the coronal one-third of the root surface the fibers are arranged horizontally. In the middle third of the root the fibers run obliquely from the occlusal surface to the alveolar bone. The apical third maintain an oblique arrangement but the fibers run apically from the cementum surface to the alveolar bone (Grant and Bernik 1972, Bartold and Narayanan 1998).

Ligament formation in teeth with and without primary predecessors differs in structure to some degree. Grant et al. (1972) found that ligament formation in deciduous teeth differs from succedaneous teeth. The processes are not unique as both classes of teeth follow the same stages; however the timing of development is delayed for secondary teeth. During the preeruptive stage the succedaneous premolar shows only a few fiber extrusions from the cementum and no fibers are apparent from bone. Most of the PDL space is filled with loose collagenous elements. The permanent molar has well defined predentogingival and alveodental fibers that extend between bone and cementum. Upon eruption the succedaneous tooth only shows organized dentogingival, alveolar crest and horizontal fibers, leaving the rest of the ligament in developing stages. The molar demonstrates continuous principal fibers at the eruption stage. During initial occlusal contact the succadeaneous premolar demonstrates organized and continuous alveolodental fibers for the coronal two-thirds of the root. The principal fiber formation is still progressing in the apical one-third. The molar exhibits continuous periodontal ligament fibers. During full occlusal function both the molar and premolars exhibit classically aligned and thickened PDL fibers. So, although the developmental timing differs, after eruption and a period of occlusion the fibers in secondary and primary teeth thicken and become indistinguishable from each other (Grant et al. 1972).

Composition

A healthy, functioning PDL contains multiple cell types. Commonly the PDL is composed of fibroblasts, endothelial cells, sensory system cells, bone associated cells and cementoblasts. Collectively these cells act to sense applied physical forces and respond to them by maintaining periodontal width and preserving cell viability (McCulloch et al. 2000). The main component of the PDL is fibroblasts. In rodents, fibroblasts make up 35% of the volume of PDL space, in sheep closer to 20% and in humans approximately 25-30% (Beertsen 1975; Berkovitz and Shore 1995, McCulloch et al. 2000).

Matrix Morphology

The PDL is mostly composed of oriented bundles of collagen fibers called principal fibers. The individual collagen fibers within the bundles are approximately 55nm in diameter. These fibers are small compared to the 100-250nm collagen fiber diameters observed in tendons. A large diameter is generally attributed to older fibers; the small diameter in PDL fibers is likely due to a high rate of collagen turnover (Sloan and Carter 1995). Adult human fibrils are slightly thicker than rodents and several other mammalian species indicating a variable rate in PDL turnover (Sloan and Carter 1995). The fibers of the PDL tend to be wavy on their course from cementum to bone, which allows for movement of the tooth (Melcher and Eastoe 1969). Recent studies have found no functional link between individual fiber thickness and mechanical response in bovine PDL (Pini et al. 2002).

The PDL's collagen fiber bundles are primarily composed of interstitial collagens I and II. These collagens form banded fibrils (Bartold and Narayanan 1998). Collagen V is also associated with the fibrils, however it is found either in the spaces between fibril bundles or within the core of the fibrils (Huang et al. 1991). Several other minor collagens are also found in the PDL such as collagens VI and XII as well as other extracellular matrix proteins including some proteoglycans, fibronectin and plycoprotiens (Karring et al. 1993, Beertsen et al. 2000, McCulloch et al. 2000)

Of principal importance to the function of the PDL are the attachment sites to tooth and bone. Sharpey's fibers are the extensions of the principal fibers of the PDL into the hard tissues – cementum and bone. Once embedded into the wall of the alveolus or the cementum Sharpey's fibers calcify and are associated with noncollagenous proteins commonly found in bone and tooth cemetum (McCulloch et al. 2000). Sharpey's fibers tend to be longer on the tension side of the PDL (appositional side) suggesting interstitial fiber growth where the bundles are integrated into the surrounding bone (Beertsen et al. 2000). Sharpey's fibers are also associated with high levels of ostepontin and bone sialoprotien; this close association could possibly provide advantageous physical properties to the hard tissue – soft tissue interface. Bone remodeling severs Sharpey's fibers as old alveolar bone is replaced by new bone. The association with noncollagenous proteins would allow for continuous embedding of PDL fibers into the alveolar wall (McCulloch et al. 2000).

Fiber bundles appear to have constant orientation relative to their location. Accordingly, PDL fibers are generally been classified into categories based on orientation: gingival, crestal, horizontal, oblique and periapical fibers (Melcher and Eastoe 1969; Sloan 1978; Bartold and Narayanan 1998). Alveolar crest fibers run from the cementum of the tooth in an apically slanted direction and towards the alveolar crest. Horizontal fibers run in an occlusally slanted direction from the cementum covering the apical two-thirds of the root to the alveolar bone. Oblique fibers run in an occlusally slanted direction from the cementum covering the apical two-thirds of the root to the alveolar bone. Periapical fibers radiate from the cementum of the apex to the alveolar bone. Not all fibers bundles present in the PDL insert into alveolar bone. Gingival fibers run from the most cervical cementum into the lamina propria of the gingiva. Circumferential fibers originate in the most cervical cementum and run horizontally around the root and insert into the lamina propria of the gingiva. Transseptal fibers run from the cementum of one tooth, cross the interdental septum and insert into the cementum of the adjacent tooth. In multi-rooted teeth interradicular fibers run from the interradicular septum into the cementum (Moss-Salentjin and Hendrick-Klyvert 1990).

These classification schemes have been challenged especially for animals with continually growing teeth (Sloan and Carter 1995). The organization of PDL collagen fibers in general is closely aligned with *in vitro* load characteristics (Viidik 1980), indicating that the morphology PDL is modified by applied loads (Komatsu et al. 1998). This suggests a close link between PDL morphology and ability to withstand or buffer applied forces.

Matrix Remodeling

The maintenance and remodeling of the PDL collagen fibers and Sharpey's fibers require a coordinated synchronized action of multiple cell types and signaling pathways. Non-human animal studies have indicated that there is an extremely high turnover rate for PDL matrix proteins (Sodek 1989). Turnover differs from remodeling in that turnover constitutes no change in the structural organization of the tissue while remodeling implies positional or functional changes in the tissue. Both processes occur simultaneously and the speed of remodeling seen for the PDL attributes to the unique character suite that is part of its adaptability (Sodek 1989).

The vascular elements and extracellular matrix proteins of the PDL function to allow teeth to adjust their position within the tooth socket while remaining firmly attached. During remodeling the collagenous mesh that stretches from bone to cementum must be rapidly broken-down and synthesized. Studies of continuously growing molars had indicated that remodeling of the ligament is restricted to the mid-region where fibers from the bone and teeth intermingle, the "intermediate plexus" (Sicher 1942). More recent work in teeth with limited eruption found remodeling activity throughout the PDL from cementum to bone (Rippin 1976). Fiber systems are broken down to adapt to changes in the position of teeth. The morphology of the PDL, as a stretched out net, allows for regional breakdown of the meshwork without compromising the integrity of the tissue on the whole (Beertsen et al. 2000, McCulloch et al. 2000).

Central to the remodeling activities are the PDL fibroblasts. The fibroblasts are responsible for the formation and remodeling of periodontal fibers. The process of collagen breakdown is controlled intracellularly through phagocytosis. Early work suggested that collagen breakdown was mediated by extracellular expression of the enzyme collagenase, but an intracellular mechanism allows for the precise removal of collagen fibers (Beertsen et al. 2000).

Sensory mechanisms and receptors in use by the PDL are still under investigation. PDL cells have mechanisms to respond directly to mechanical forces by activation of mechanosensory signaling systems. These systems include stretch-activated adenlyate cyclase ion channels and changes in cytoskeletal organization. These immediate responses to mechanical force result in the production of intracellular messengers. After physical osteoblast stretching, increased inositol phosphate concentrations have been observed. Also, intracellular calcium ion concentration oscillations have been observed in PDL cells in response to substrate tension (Jones et al. 1991, Carvalho et al. 1994). In vitro work has demonstrated an increase in prostaglandin release in tension areas of the PDL (Yeh and Rodan 1984).

Longer term responses to loading include altered collagen synthesis, promotion of collagenase activity, stimulation of cell division and release of transforming growth factor-ß (Leung et al. 1976, Curtis and Seehar 1978, Jones et al. 1991, Carvalho et al. 1994; Klien-Neuland et al. 1995). These studies indicate that there are many potential routes that may lead to direct or indirect influence of the PDL on bone remodeling, however the exact pathway is still unclear. There is a reciprocal, dynamic relationship between activation of membrane-localized cation-permeable channels and cytoskeleton structure. Also, the fibroblasts and osteoblasts that populate the PDL have the required signaling and effector mechanisms to sense and respond via remodeling to physical force (Beertsen et al. 2000). For example, intermittent pressure application to PDL cells increases bone resorption (Saito et al. 1991). Although the exact signaling and cellular mechanisms responsible for PDL behavior is unclear, it is evident that the PDL is a requirement for rapid remodeling of bone via application of physical forces to the teeth.

Proprioception

One of the main functions of the PDL in the masticatory cycle is to provide sensory feedback during chewing. Humans are capable of detecting the presence of very small particles between the occlusal surfaces of teeth. The teeth also can serve as an excellent judge of material properties. There are proprioceptive sensors in the PDL that provide sensory information about how fast and how hard to bite (Hannam 1982). While there are other concentrations of sensory fibers in the oral cavity the PDL plays a substantial role in providing unconscious feedback for the masticatory system.

Lund and Lamarre (1973) anesthetized patient's teeth and found a 40% reduction in bite force applied, indicating that PDL proprioceptors are important in the control of bite force. Cathelineau and Yardin (1982) found that patients with advanced periodontal disease had an increased threshold in detection of vibrations sent through the teeth, most likely the result of inflammation of the ligament. Edentulous individuals still display proprioceptive abilities but dentate individuals have the ability to produce much higher interocclusal forces and differentiate between fine particles with higher accuracy (Hannam 1982). Edel and Wills (1975) and Williams et al. (1986) found that individuals with loss of periodontal bone but no tissue inflammation had no significant differences from a control population in perception of axially applied forces. Williams et al. (1985) anesthetized teeth and TMJs to measure differences in inter-incisor bite force. They found discrimination of bite force was significantly impaired only when the teeth were anesthetized, again implicating the PDL in a sensory feedback role. These studies indicate feedback mechanisms of the PDL play a significant role in the proper mechanical function of the masticatory system by assisting in particle identification and maintaining appropriate force application.

Mechanical Profile

The PDL is subjected to forces during mastication, speech and orthodontic tooth movement. Because it is confined in the alveolus and loads applied by tooth-tooth contact during mastication are not purely axial the PDL experiences both compression and tension during a regular loading scheme (Pini et al. 2002). The mechanical strength of the PDL is derived from the molecular structure of the type I collagen and its arrangement into fibers (McCulloch 2000). There appears to be regional variation in the biomechanical properties of the PDL, variation that may depend on the developmental stage of the collagen fibers as well as arrangement, diameter and density of the collagen fiber bundles (Komatsu et al. 1998). The PDL is also highly vascularized, a trait that attributes to its viscoelastic behavior. Matsuo and Takahashi (2002) found that the blood vessels in the PDL may contribute to "shock absorber" behavior of the PDL, cushioning the alveolus from the occlusal load. Exact behavior of the PDL during mastication is still somewhat unclear, and in vivo research on human subjects is difficult to come by. Most experimental studies of PDL response in vivo has been done on rodents. Interspecies variation in the mechanical properties of the PDL has been attributed to differences in the width and waviness of the PDL as well as the strength and stiffness of the species specific periodontal collagen fibers (Komatsu et al. 1998).

A primary role of the PDL is to act as a medium of force transfer during mastication. The PDL displays some unique mechanical properties that may be ascribed to loading regimes commonly observed in mastication. Despite some uncertainties about detailed behavior and remodeling effects of the PDL, research has demonstrated broad cross-species trends in PDL behavior.

One of these trends is that the PDL exhibits viscoelastic behavior, where the fluid component of the tissue modifies the action of the fibers in withstanding transmitted loads. As increasing levels of force are applied to the tooth the force-displacement curve is found to be non-linear. The initial resistance is low but as the force is increased the resistance increases until at high levels the additional displacement is very small (Wills et al. 1978). Ralph (1982) found that in tension human PDL exhibited viscoelastic behavior in both extrusion and intrusion and suggested that the close interaction of PDL fibers and capillaries may produce a closed fluid system in the PDL space which would allow for distribution of large masticatory loads to the alveolar wall. Results of experimental extrusion and intrusion of human teeth produced stress-strain curves that had distinct toe and linear regions indicating a nonlinear response to load (Toms et al. 2002a, 2002b). The response varied according to the age of the PDL donors, region of the PDL measured and direction of load application. Toms et al. (2002a, 2002b) recommend the inclusion of these unique viscoelastic properties into computer simulations of orthodontic tooth movement.

Pini et al. (2004) found that bovine PDL exhibited non-linear and time-dependant mechanical behavior. They also found that the maximum tangent modulus, strength, strain and strain energy density all increased with depth of location for the incisor and for the molar (except for the apical region of molars), supporting earlier work that the properties of the PDL vary along the length of the root. Their results indicate that the PDL should be viewed as a biphasic material, similar to articular cartilage, in future experimental work; and that the compressive behavior of the PDL is highly influenced by interstitial fluid flow. Martin et al. (1998) established that the cartilage and tendons are stiffer at higher load magnitudes, indicating that there is no need to model them as biphasic tissues for normal physiological loading (Carter and Beaupre 2001).

In conjunction with viscoelastic behavior the PDL also exhibits a distinct toe region, a loading region where relatively large increments of strain occur with low changes in stress. This is counter to early work assuming a linear response to load. Komatsu (et al. 1998) found that the maximum functional shear stress exerted on the PDL is approximately 0.335 MPa, a value which corresponds generally to the upper limits of the toe region. These findings indicate that the PDL naturally operates within the toe region and that the linear stress-strain curve would not be readily observable in natural loading scenarios. The toe region is generally attributed to fiber reorganization with no significant change in strain. With further displacement the load continues to increase non-linearly, probably the result of principal fibers aligning with the direction of force application. Once the fibers are uncoiled a linear stress-strain curve is usually observed and attributed to the mechanical stretching of the fibers themselves (Pini et al. 2002).

Rate Dependency

Parfit (1960) found that experimental intrusion of teeth was affected by the rate of load application and the interval between loads in macacques. Wills et al. (1978) using two rates of load application, 4 $Nsec^{-1}$ and 12 $Nsec^{-1}$ and human PDL samples, found that slow application of force result in more tooth movement than rapid force application. Wills suggests that the slower movement allows the extracellular fluids within the PDL to move whereas a rapid application would trap them causing restriction of tooth intrusion. They suggest that at high loading rates all of the components of the PDL, both fibers and extracellular fluid, remain in place and act as a single unit to transmit loads.

The values for rate dependency of ligament load-responses under monotonic loading vary widely. Some rabbit knee ligament data indicate minimal rate dependency (Woo et al. 1990) while rat PDL appears have a higher level of rate dependency. Komatsu and Chiba (1993) examined the mechanical responses of rat PDL at molar and incisor locations at various load rates. Mechanical strength, stiffness and toughness were greater for the molar than the incisor ligament indicating that the molar ligament probably has a different fiber arrangement than the incisal ligament. However; both the incisor and molar ligament demonstrated mechanical differences based on loading velocity. Their primary finding was that stress levels were reduced at lower loading velocities. Pini et al. (2002) found no rate dependency of stress as a function of strain in bovine PDL push-pull tests. However, in similar experiments with torsion using human PDL there appeared to be a rate dependency for hysteric effects (Daly et al. 1974).

Hysteresis

Hysteresis is the lag in the response of a material to the application or removal of a load. It is a key characteristic of viscoelastic materials and a measure of the ability of a material to store and dissipate energy. Purely elastic materials produce identical load-deformation curves for the loading and unloading phases. In viscoelastic materials the area under the loading curve is greater than the area under the unloading curve. The difference between the two is a measure of hysteresis, the energy lost during the load cycle (Martin et al. 1998). Mechanical push-pull tests on bovine PDL suggest the PDL response in the range of physiological compression exhibits high levels of hysteresis (Pini et al. 2002). Hysteresis resulting from tension is due to the uncoiling and friction between principal fibers (Pini et al. 2004). Some of the stiffness in PDL response to small loads has been attributed to its high level of vascularization, which also may account for some level of the hysteric behavior seen in the PDL. Additionally, the level of vascularization may affect the accuracy of small loads (0.5N-3.5N) used *in vitro* studies due to lack of a maintained circulatory system (Wills et al. 1976, Pini et al. 2002).

Understanding of the PDL's role in force transfer and sensory feedback has grown substantially. Although there are still many aspects that are unknown it is clear that the PDL is in integral part of the masticatory system and its organization and functional role should not be overlooked. Investigations of the PDL can potentially inform a myriad of masticatory inquiries from processes of bone metabolism to biomechanics and masticatory physiology. The PDL is a vital component of the masticatory system and furthering knowledge on the physiological and biomechanical fronts are imperative to a clearer understanding of masticatory system as a whole.

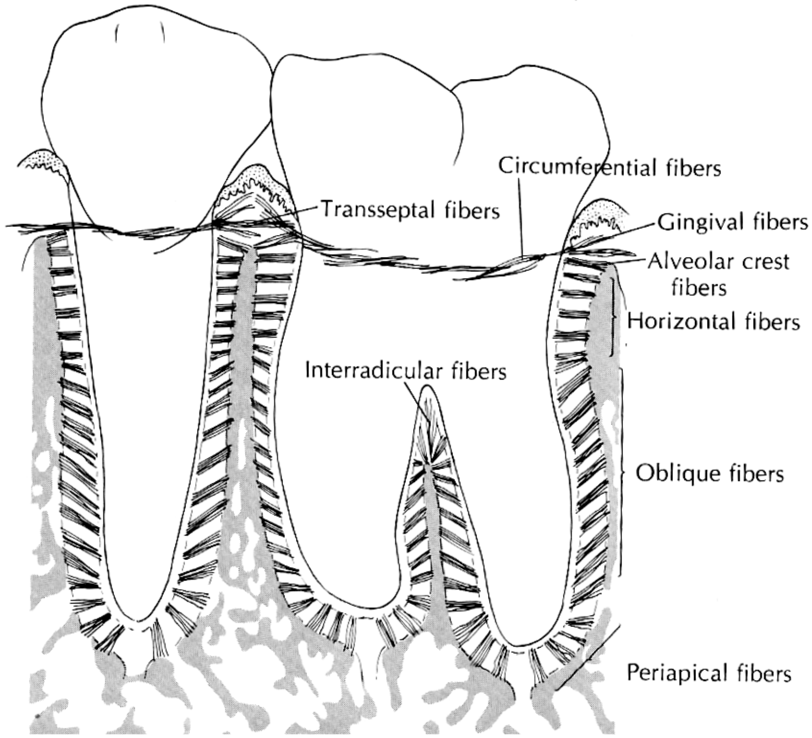

Figure 2-1. Schematic of PDL fiber orientation. Adapted from Moss-Salentjin and Hendrick-Klyvert (1990).

CHAPTER 4
FINITE ELEMENT MODELS

Finite element modeling (FEM) is commonly used in engineering applications to determine patterns of fluid flow, stress, strain, heat transfer and other problems requiring complex mathematical computations. Modeling in biomechanics offers an alternative to invasive techniques commonly used to explore the mechanics of the face (Korioth et al. 1992). It is a powerful tool and is growing in popularity, but is only useful if applied appropriately. One goal of this research project is to provide a guideline for the importance of PDL inclusion in future models of the skull. The PDL is commonly included in cranial and mandibular models created for dental and orthodontic applications (Wilson et al. 1994, Rees 2001), but is often ignored in anthropological and comparative models (Marinescu et al. 2005, Richmond et al. 2005, Strait et al. 2005).

Overview and Requirements of Modeling

Finite element analysis (FEA) can be subdivided into three main processes: building the model, loading the model and validation of the model. All three steps play in integral role in the information produced by the model.

Building a Model

The construction of a model consists of recreating the specimen geometry and then outfitting that geometry with the appropriate material properties within the FEM program. The importance of accurately representing specimen morphology and material properties in the model can not be stressed enough. Unless this stage of the modeling is completed accurately and thoughtfully the resulting predictions will not be meaningful.

Capturing geometry

The starting place of model generation is digitally capturing the geometry of the specimen to be modeled. Traditional methods include laser scanning or digitally tracing the outline of a specimen. More recently computed tomography (CT) and magnetic resonance imaging (MRI) data have been used to generate the appropriate specimen geometry. CT scans are appealing because of high resolution and ease of inputting the data into a computer, but it is not without its problems. There are complex issues involved in appropriately delineating the specimen boundaries (Ryan and van Rietbergen 2005). In addition the use of detailed CT data can result in a model with geometry so complex that the computing power needed to process the model is unavailable. Use of CT and MRI data is preferred to methods that only account for the external shape of the specimen (i.e. laser scanning or digital tracing). Internal geometry has an effect on structural properties of the model as a whole so methods of geometry capture should be able to account for differences in internal structure. Creating the geometry is usually the most time consuming portion and assumptions made at this stage can have a drastic effect on the results (Ross 2005).

Once the gross geometry has been captured it must be subdivided into smaller, geometrically simple areas (the so-called "finite elements"). These units of area (or volume) are connected at their vertices via points referred to as nodes. The collection of connected elements is the mesh. Displacements are determined with reference to nodes. Depending on the dimensionality of the problem there are several different shapes of elements that can be utilized. (Beaupre and Carter 1992).

A criticism of FEA is that it is difficult to incorporate morphological variation. It is impractical to productively model population-level variation. Models are generally created using scans of a single specimen, so assumptions must be made as to the representative nature of the specimen. Because generating a model can be exceedingly time consuming, usually only one specimen is modeled and all loading scenarios are run on the same individual (Ross 2005).

Attributing material properties

As FEA has become more commonly used in biomechanical studies the difficulties in modeling irregular shapes and anisotropic materials have become more pronounced. Bone is irregularly shaped, has variable densities and elastic moduli and exhibits viscoelastic properties. In order to create an accurate model one must take into account the variation in the material properties unique to the areas of bone being modeled. To account for all the changing properties would require a crippling time commitment. To circumvent the problem some assumptions are commonly made in creating models:

- It is commonly acceptable to use extant animal properties for extinct animal modeling (Rayfield 2005)
- In some cases isotropy is assumed and material properties are established using nanoindentation studies of comparable bone (Silva et al. 2005)
- Assuming isotropy and using published average values for material properties is also common (Thomason 2005)

Recent studies have indicated that accurate material properties are essential to a good model. Both Marinescu et al. (2005) and Richmond et al. (2005) found that different inputs of material properties profoundly affected the resulting strain predictions. In general an accurate model relies on detailed geometry and relatively realistic assignment of material properties.

Loading a Model

Once an appropriately detailed model has been constructed the next step is to “load” and “restrain” it mathematically. Creating an informative loading scenario requires accurate data on the external forces acting on the specimen. Although it can be difficult to gather comprehensive data on the forces anticipated during a loading event in vivo, the resultant strains are necessarily completely dependent on the programmed loading environment. Marinescu et al. (2005) found that small changes in the direction of load application and restraint had significant effects on predicted strain. While it is certain the predictions of a model are contingent on the details of construction and loading, the relative importance of loading conditions, material properties, geometry and constraints are still under investigation (Dar et al. 2002; Marinescu et al. 2005; Ross et al. 2005; Richmond et al. 2005, Strait et al. 2005).

Validation

FEA produces a prediction of how the modeled object would react to the modeled load. The resulting prediction should be compared to in vitro or in vivo data to ensure the validity of a prediction and that it lies within expected parameters. Currently, most papers provide modeling properties but do not actually validate (Richmond et al. 2005). Few validations are thorough and accurate; some notable exceptions are Coleman et al. (2002, 2003); Kotha et al. (2003) and Srinivasan et al. (2003). In each of these studies modeled predictions were painstakingly validated with relevant in vitro and in vivo data. Models of human or extinct bone behavior can be difficult to directly validate as extensive in vivo work has not been done with humans and cannot be done with extinct organisms. In these cases cross-species comparisons are commonly applied.

Review of Relevant Studies

It is unclear how much of an effect simplifications in model creation can have on the resultant predictions. Inclusion of the PDL is common in orthodontic models and those incorporating occlusal loads, but simplification or removal of the PDL is often observed in comparative models. It is in this context that the current experiments have been proposed. They are an attempt to ascertain the in vitro results of the PDL being removed from the force transfer scenario in occlusal load applications. The role of the PDL in modeling scenarios has generally increased as the use of finite element modeling has grown.

Knoell (1977) generated a model of a human mandible to investigate the biomechanical response of the tooth support region to occlusal loads. Knoell's model did not include the PDL but it was determined that biomechanical response is highly localized to the region of tooth support and therefore a model of the region near load application requires careful simulation of the surrounding anatomical structure, including the PDL. Atmaram and Mohammed (1981) created a model to investigate the difference between representing the PDL as a continuous structure and breaking it up into a fibrous structure, similar to its actual anatomical arrangement. They determined that the type of PDL modeled had a significant effect on alveolar stresses.

In the human mandible model generated by Korioth et al. (1992) the periodontal region was divided into three regions, each with its own PDL properties. Material properties of the PDL vary along the root (Mandel et al. 1986); in an attempt to avoid oversimplification Korioth et al. (1992) modeled the PDL with varying elastic moduli. Wilson et al. (1994) produced a model to investigate the role of orthodontic stresses in alveolar bone remodeling. Considering the PDL plays a role in bone remodeling of the tooth region, they felt it was pertinent to include PDL material properties and morphology in model construction. Similarly, in a study of orthodontic forces on teeth and surrounding tissues Jones et al. (2001) included the PDL in their model since the PDL is the mediator of tooth movement. Rees (2001) also stresses the importance of including the PDL when undertaking finite element studies of teeth. It is generally accepted that models of the tooth and the immediate area are unreliable and inaccurate without inclusion of PDL material properties (Khera et al. 1988, 1991, Goel et al. 1990, 1992, Korioth and Hannam 1994).

Despite such studies, some models are generated without inclusion of PDL material properties. Hart et al. (1992) created a mandible model investigating isometric biting and the role of missing teeth in the dissipation of stress. The tooth-bone interaction was modeled using the stiffness of cortical bone with no interface layer. The authors acknowledge that lack of a PDL in their model inhibits its capability to accurately represent the stress environment in the alveolar region. Another finite element model of the mandible without a PDL was created by Marinescu et al. (2005). Their study assumed that inclusion of the PDL in modeling was unnecessary as the teeth the PDL do not significantly contribute to the structural stiffness or strength of the mandible during non-occlusal loading. To investigate their assumption two models were created, one edentulous with root spaces modeled as gaps and another dentate model with no PDL. After application of occlusal loads it was found that the dentate model without PDL was overly stiff whereas their edentulous model produced a more reasonable strain pattern. Modeling of the mandible and teeth with no PDL interface produced an overly stiff model and an inaccurate strain pattern.

Richmond et al. (2005) included a finite element model lacking a PDL as a case study in their review of modeling practices. While their model was primarily concerned with muscle forces the study did contain occlusal restraints and loads, indicating that results may not have been accurate due to oversimplification of dental structures. This is especially pertinent to their model of the palate as it is intimately related to dental alveolar morphology.

Overall it seems that the field-specific areas of cranial and mandibular finite element literature have chosen to deal with the inclusion of the PDL differently. While the conclusion that models of non-occlusal forces should not need to include the PDL may not be invalid, it seems an oversimplification to model skull behavior without anticipating the complex structural context of in vivo loads. In addition to muscle force, loads come from mastication forces. The goal of these experiments is to help elucidate the degree of detail required in modeling of dental and alveolar structures for inclusion in cranial finite element models.

CHAPTER 5
MATERIALS AND METHODS

Experiments were done to determine the effect of the PDL on alveolar bone strain recordings of rate dependent behavior, hysteresis and overall load response. A detailed understanding of the role of the PDL in these load scenarios helps to inform its inclusion in masticatory finite element models and more generally its function in the mastication system.

Experimental Methods

Experimental strain analysis was performed on five previously frozen, fresh *Sus domesticus* crania. Due to availability, non-human tissue has frequently been used to conduct PDL experiments. The pig model has proven to be a reliable substitute for primate experimental material due to gross similarities in their molar form, masticatory cycle and interaction between tooth surface and food particles (Herring 1976 and Popowics et al. 2001). Additionally Pini et al. (2004) found no significant difference in PDL microstructure between herbivores and omnivores. Use of previously frozen ligament stored at -10ºC to -20ºC and short-term saline storage have not been found to produce a detectable difference when compared to fresh ligament (Viidik et al. 1965, Chiba et al. 1982, Woo et al. 1986, Pedersen et al. 1991, Pini et al. 2002)

Intact PDL Experiments

The aim of the first stage of the experiment was to measure strain dissipation in the cranium with an intact PDL, accordingly, all flesh was removed and the cranium was cleaned manually with scissors and scalpel using no solvents or methods that would risk the integrity of the PDL. Between experiments the cranium was wrapped in saline-soaked cloth and refrigerated.

Following removal of soft tissues the back of the cranium was mounted in an epoxy block beginning 1.5cm posterior to the superior orbital margins. The epoxy block was used to secure the cranium to the test stand with a mechanical fixture. The mechanical fixture restrained the block via two graphite bars secured to the test stand by turnbuckles and bolts. The cranium was immobilized with the occlusal surface facing superiorly to allow access for load application. Surface bone strain data were collected from three rosette strain gauges (FRA-1-1L rectangular rosette(stacked), 1 X 0.1 mm gage length, 4.5 mm diameter backing, Texas Measurements, College Station, TX) bonded to predetermined sites of interest on the maxilla.

In preparation for bonding the periosteum of the maxilla above the Ldm^1 was removed and the bone was degreased with isopropyl alcohol, etched with a 5% solution of phosphoric acid and neutralized with a 0.02% ammonium hydroxide solution. Two gauges were bonded to the left maxilla at the margin and at the root apex level of the alveolar bone of the Ldm^1. A gage was also bonded to the left mid-facial region just posterior of the infraorbital foramen. Tests were performed using MTS 858 MiniBionix Test System (Eden Prairie, MN) in displacement mode with a 407 controller. A force transducer affixed to the test system actuator permitted simultaneous recording of loads and displacements during occlusal force application.

The cranium was subjected to a vertically applied occlusal load at the Ldm^1. Since the PDL's response to load varies according to application rate (Komatsu and Chiba, 1993; Chiba and Komatsu 1993; Pini et al., 2004) and the PDL distributes applied forces to the contiguous alveolar bone (Beertsen et al. 2000) the aim of these experiments was to determine if the rate dependant behavior of the PDL could be observed in surface strain patterns in the alveolar and facial bone. Loads were applied at exponentially increasing rates beginning with .2mm/sec and ending with 3.2 mm/sec for a total of five experimental iterations. Within each iteration the rate of load application was held constant. Strain recording commenced at 200N corresponding to the lower range of maximum occlusal forces in pigs (Popowics et al., 2001). Data were recorded at 100 points per second for 100s or until the load reached 800N. Loads past 800N were not analyzed as at higher loads deflection of the restraints was noted. Additionally the 800N maximum falls near the maximum biting force for humans which has been recorded at around 720 N, Atkinson and Ralph 1976).

Cyanoacrylate Experiments

The second aim of this study is to compare the dissipation of strain in the cranium via an intact PDL to that of a direct tooth-bone interface. After completion of experiments utilizing the intact PDL, the Ldm[1] was excised and all remaining ligament was mechanically destroyed. Picton (1991) found that disrupting the PDL via extraction was enough to prevent the fiber network from developing tension during reloading, thus precluding normal PDL mechanical response. After PDL removal, the molar was then glued back into position using cyanoacrylate adhesive and allowed to cure for 24 hours before testing resumed, repeating the previously outlined experiments. Cyanoacrylate was chosen over other adhesives because its Young's modulus is consistent with that of alveolar bone (Martin et al. 1998, Loctite 2001).

Dynamic Loading

Although the main focus of this paper are the rate experiments, dynamic load experiments were also conducted on both intact PDL and cyanoacrylate bonded teeth for three specimens. This set of experiments was conducted to evaluate the behavior of alveolar bone in a dynamic load scheme similar to that of mastication. Tests were performed using MTS 858 MiniBionix Test System (Eden Prarie, MN) in displacement mode with a 407 controller immediately at the conclusion of the rate experiments described above. The specimens were loaded using a sine curve loading program beginning with .5Hz and exponentially increasing to a maximum frequency of 2.0Hz. This range approaches the physiological chewing cycle of pigs, generally 2-3Hz (Rafferty and Herring 1999). Additionally the displacement setting was altered to achieve varying loads and load rates. The displacement span was increased in increments of 2mm (from 2mm to 12mm). Due to an inability to effectively prevent specimen creep during cyclical loading only one specimen (Specimen 5) was successfully tested.

Analytical Methods

Strain gauge signals for all experiments were processed through Measurements group 2120 conditioner/amplifiers (Raleigh, NC) and were recorded and analyzed using Superscope II software (GW Instruments, Somerville, MA). Strain gauge signals were converted to maximum principal strain (ε_1), minimum principal strain (ε_2), shear strain (γ) and principal strain ratio ($\varepsilon_1/\varepsilon_2$) prior to analysis. During occlusal loading the angle of load application and alveolar morphology produce direct shear stresses (Demes et al. 1984), accordingly shear strain values are of primary interest to this study. As the goal of this experiment was not to reproduce the in vivo strain environment, the absolute strain values are not of direct concern. The data of interest to this study are 1) the relative differences in strain for the five load rate scenarios, 2) the comparison of intact PDL and direct tooth-bone interface strain distribution patterns, 3) the relative differences in strain for the three gage locations and 4) observations of hysteresis during cyclical loading.

To evaluate the effects of load rate the slope of a load-deformation curve was generated for each specimen at Gage 1 for all load rates for the intact PDL. An ANCOVA was used to determine the statistical significance of variation in slope between the five ramp-load iterations within each individual. To compare the effects of PDL removal, slopes from a load-deformation curve and maximum shear strain values were compared between the intact PDL and bonded tooth experiments. Gages 1-3 at all rates were compared. These data were evaluated for broad trends in behavior.

The rate at which strain changes as a function of distance by gage location, was measured. As these three gages were in a straight line from the loaded tooth they provide a means for evaluating the dissipation of strain away from the load. In addition, the difference between intact PDL and bonded tooth strains and load-deformation slopes at each gage location were compared to evaluate the magnitude of the PDL effect at increasing distances. During cyclical loading shear strain values were taken for Gage 1 at the instant of load removal and load application to determine the residual strain at each load frequency. This was repeated for early and late loading cycles within each experiment to evaluate any changes in hysteresis based on number of cycles.

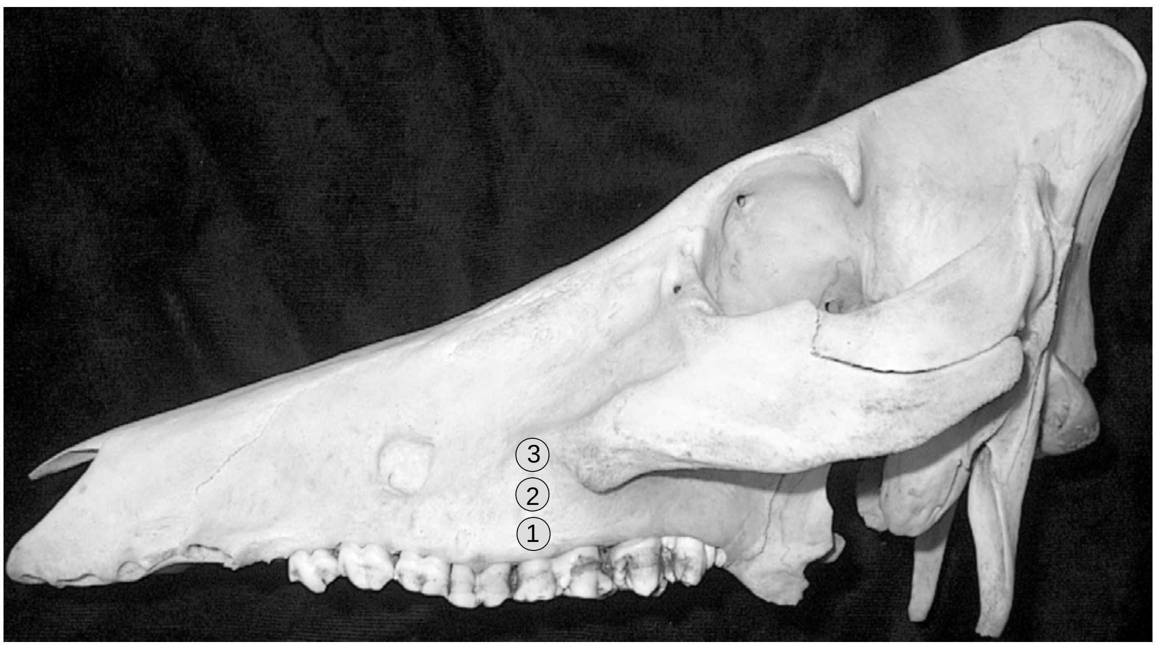

Figure 4-1. Diagram of gage placement. 1) Inferior alveolar rosette, 2) Intermediate alveolar rosette, 3) Facial rosette.

CHAPTER 6
RESULTS

Effect of Bonded Tooth

The cyanoacrylate interface produced markedly different strain magnitudes in four of five specimens analyzed. Specimen 2 only demonstrated small differences in strain for intact versus bonded tooth. These data are presented in Tables 6-1 through 6-5. The lack of response in Specimen 2 is possibly an artifact of incomplete PDL destruction during testing which will be discussed in the next chapter. In these experiments the slope of the load-deformation plot is considered a relative measure of force transfer from tooth to bone. The slope-deformation line illustrates the level of strain recorded per unit of load applied. A higher slope implies a larger bone reaction per unit load than a lower slope, indicating that the load was more effectively transferred. The maximum shear strain and slope of the load-deformation curve tended to be higher in the glued experiments for Specimen 1, but at Gages 2 and 3 only. At Gage 1 the shear strains and slope were higher in the PDL experiments. Specimen 2 exhibited consistently larger strains in the glue case, although the values of the difference are small. Specimen 3 also exhibited higher strains and higher slopes for the glued experiments. The response for Specimen 3 was more uniform, the shear strains were higher and the slopes were larger at all gage locations. The response of Specimens 1, 2 and 3 indicate that the bonded tooth transmitted more load to the alveolar bone than with the PDL present.

Specimens 4 and 5 exhibit the reverse trend. Both demonstrate higher shear strains in the PDL case, excluding Gage 2 at the lowest load rate. The slopes at all gage locations for both Specimens 4 and 5 indicate the PDL was more effective at transmitting load than the bonded tooth trials. Even though the data are somewhat inconclusive in determining which is the most effective load case (PDL or glue), four of the five specimens did display substantial differences in strain magnitude and load-deformation slope when the PDL was replaced with glue.

Location Effects

The gradient of strains for each specimen can be seen in Tables 6-1 through 6-5. With the PDL intact, strains decrease with distance in Specimen 1 and appear to increase with distance in all other specimens. In the glue case there appears to be a spike in strains at Gage 2 for Specimen 1, and a general increase in strains with distance in Specimens 2 and 3. Specimens 4 and 5 generally exhibit an increase in strains from Gage 1 to Gage 3; however Gage 2 registered the smallest strains. Overall there is a clear tendency to see higher strains apically in the intact PDL experiments.

The strain gradients created by the PDL and glued experiments were the same in two specimens. The strains were identical in distribution, differing only in magnitude. Specimen 4 produced a gradient that was essentially the same after PDL removal, the only dissimilarity was at the lowest load rates. Specimens 1 and 3 demonstrated clear changes in the pattern of strain distribution after removal of the PDL. In Specimen 1 the PDL produced higher strains at Gage 1 while the glued experiments produced higher strain at Gage 2 and the smallest strains at Gage 1. For Specimen 3 an intact PDL led to smallest strains at Gage 2 and largest at Gage 3 while the glued specimen showed apically increasing strains.

Figures 6-1 and 6-2 illustrate the differences in slope and strain magnitude between the glue and PDL load cases among the three gage locations at all rates. The general trend is a reduction in difference between the intact and destroyed PDL experiments as distance is increased. For slope differences Gage 1 is only slightly more affected than Gage 2 while the strain response at Gage 3 is markedly less influenced by tooth bonding. There appears to be a more regular reduction in the maximum shear strain difference as distance increases. As the site of interest is removed from the point of load application the effects of the PDL decrease.

Effect of Load Rate

The slope of a load-deformation plot was compared for all rates of load application for an intact PDL (Table 6-6). In all 5 specimens there was significant difference in the slope of the load-deformation curve based on rate of load application. Although there was evidence of differential load transfer among rates, no discernible pattern of rate dependency could be identified.

Cyclical Loading

Cyclical loading had the greatest effect at Gage 1, the inferior alveolar rosette. It is clear that although the load is uniformly removed the strain recovery showed some level of hysteresis while the PDL is intact at 0.5 Hz (Figures 6-3 and 6-4). After removal of the PDL, hysteresis is not as evident (Figure 6-5). Data for all frequencies is presented in Table 6-7. At a frequency of 0.5 HZ hysteresis is evident in both the PDL intact and bonded tooth cases. At 1 Hz there is evidence of hysteresis in both the intact PDL and bonded tooth, although the degree of hysteresis appears to be less than at 0.5 Hz and the bonded tooth is less affected. The strains also do not recover completely and recovery worsens as the experiments continue, illustrating some level of creep. The same pattern holds true for 2 Hz, however less hysteresis is present overall when compared to 0.5 Hz and 1 Hz, probably as a result of reduced recovery time available during higher cycling frequencies. Hysteresis was also evident at Gages 2 and 3, however the effects were reduced.

Table 6-1. Comparison of Glue vs PDL reactions for Specimen 1.

Rate (mm/s)	Gage	Maximum Shear Strain (με)		Difference (με)	Slope		Difference
		PDL	Glue		PDL	Glue	
0.2	1	748	375	373	0.798	0.416	0.382
	2	472	795	-323	0.449	0.947	-0.498
	3	369	417	-48	0.372	0.586	-0.214
0.4	1	732	371	361	0.773	0.421	0.352
	2	547	799	-252	0.586	0.493	0.093
	3	409	439	-30	0.475	0.141	0.334
0.8	1	714	357	357	0.787	0.385	0.402
	2	603	813	-210	0.67	0.952	-0.282
	3	407	442	-35	0.498	0.586	-0.088
1.6	1	727	367	360	0.809	0.395	0.414
	2	658	812	-154	0.766	0.942	-0.176
	3	409	430	-21	0.522	0.578	-0.056
3.2	1	713	351	362	0.823	0.37	0.453
	2	722	809	-87	0.843	0.957	-0.114
	3	399	442	-43	0.565	0.531	0.034

Table 6-2. Comparison of Glue vs PDL reactions for Specimen 2.

Rate (mm/s)	Gage	Maximum Shear Strain (με)		Difference (με)	Slope		Difference
		PDL	Glue		PDL	Glue	
0.2	1	287	351	-64	0.316	0.402	-0.086
	2	509	521	-12	0.594	0.639	-0.045
	3	807	780	27	0.951	0.927	0.024
0.4	1	323	348	-25	0.344	0.363	-0.019
	2	509	522	-13	0.56	0.637	-0.077
	3	804	794	10	0.952	0.941	0.011
0.8	1	334	361	-27	0.365	0.374	-0.009
	2	520	524	-4	0.585	0.628	-0.043
	3	836	800	36	0.977	0.942	0.035
1.6	1	361	379	-18	0.378	0.406	-0.028
	2	503	534	-31	0.569	0.637	-0.068
	3	837	812	25	0.97	0.952	0.018
3.2	1	384	400	-16	0.403	0.435	-0.032
	2	490	549	-59	0.553	0.658	-0.105
	3	834	821	13	0.964	0.969	-0.005

Table 6-3. Comparison of Glue vs PDL reactions for Specimen 3.

Rate (mm/s)	Gage	Maximum Shear Strain (με)		Difference (με)	Slope		Difference
		PDL	Glue		PDL	Glue	
0.2	1	296	300	-4	0.389	0.321	0.068
	2	220	450	-230	0.204	0.454	-0.25
	3	519	742	-223	0.571	0.878	-0.307
0.4	1	237	309	-72	0.283	0.377	-0.094
	2	191	492	-301	0.219	0.545	-0.326
	3	465	771	-306	0.547	0.922	-0.375
0.8	1	242	355	-113	0.297	0.443	-0.146
	2	195	522	-327	0.232	0.581	-0.349
	3	483	799	-316	0.593	0.959	-0.366
1.6	1	251	381	-130	0.311	0.483	-0.172
	2	214	543	-329	0.266	0.616	-0.35
	3	526	827	-301	0.646	1	-0.354
3.2	1	171	398	-227	0.232	0.501	-0.269
	2	296	566	-270	0.443	0.652	-0.209
	3	583	851	-268	0.806	1.039	-0.233

Table 6-4. Comparison of Glue vs PDL reactions for Specimen 4.

Rate (mm/s)	Gage	Maximum Shear Strain (με)		Difference (με)	Slope		Difference
		PDL	Glue		PDL	Glue	
0.2	1	403	158	245	0.414	0.054	0.36
	2	236	291	-55	0.284	0.141	0.143
	3	499	464	35	0.591	0.455	0.136
0.4	1	413	208	205	0.43	0.255	0.175
	2	252	214	38	0.313	0.16	0.153
	3	528	425	103	0.623	0.475	0.148
0.8	1	400	296	104	0.435	0.391	0.044
	2	264	146	118	0.326	0.094	0.232
	3	532	401	131	0.636	0.451	0.185
1.6	1	369	321	48	0.412	0.402	0.01
	2	277	122	155	0.344	0.095	0.249
	3	528	417	111	0.65	0.459	0.191
3.2	1	367	332	35	0.372	0.409	-0.037
	2	268	113	155	0.344	0.116	0.228
	3	532	411	121	0.635	0.479	0.156

Table 6-5. Comparison of Glue vs PDL reactions for Specimen 5.

Rate (mm/s)	Gage	Maximum Shear Strain (με)		Difference (με)	Slope		Difference
		PDL	Glue		PDL	Glue	
0.2	1	813	387	426	0.935	0.489	0.446
	2	471	224	247	0.56	0.231	0.329
	3	1024	763	261	1.245	0.975	0.27
0.4	1	821	413	408	0.934	0.484	0.45
	2	499	231	268	0.623	0.271	0.352
	3	1069	778	291	1.31	0.98	0.33
0.8	1	823	424	399	0.927	0.493	0.434
	2	502	234	268	0.609	0.272	0.337
	3	1064	786	278	1.3	0.987	0.313
1.6	1	830	438	392	0.936	0.513	0.423
	2	499	234	265	0.604	0.252	0.352
	3	1052	790	262	1.29	0.97	0.32
3.2	1	869	447	422	0.985	0.515	0.47
	2	466	248	218	0.561	0.261	0.3
	3	1022	808	214	1.237	0.972	0.265

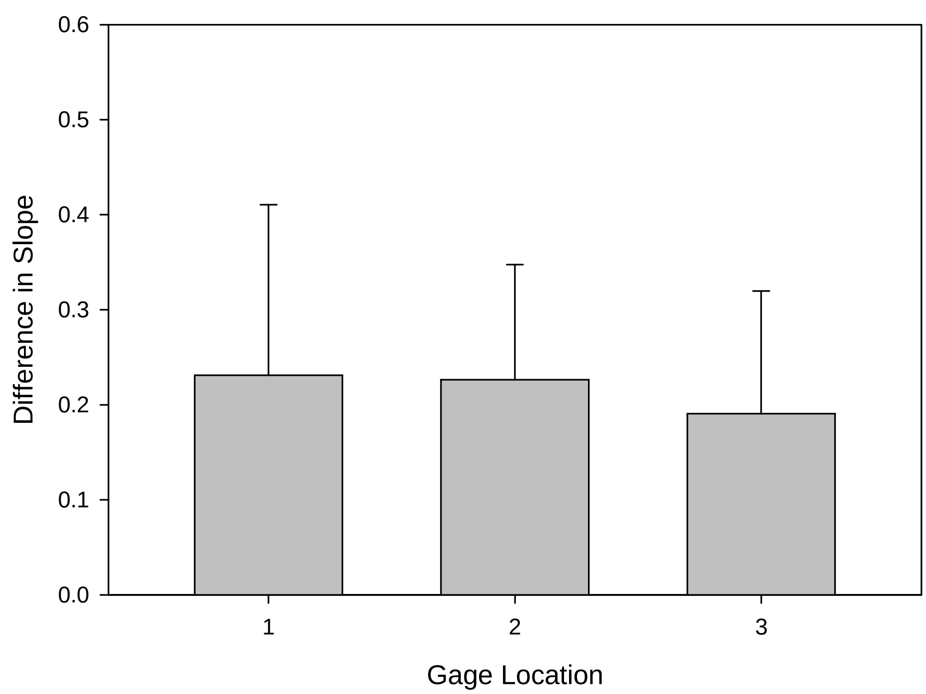

Figure 6-1. Difference slope between PDL and glue load cases at each gage location. The effect of the PDL decreases with distance from point of load application.

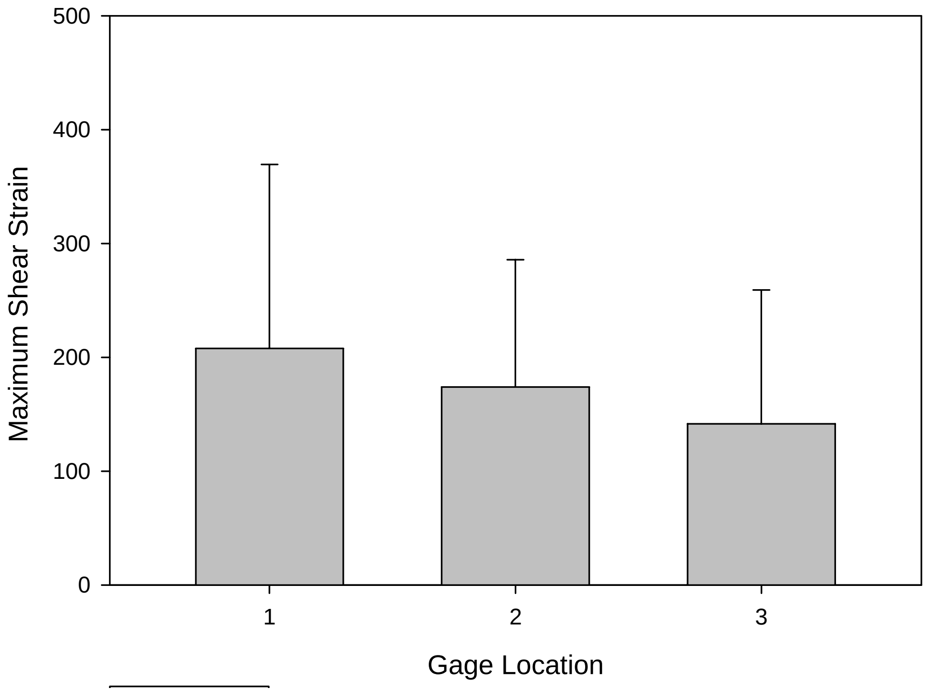

Figure 6-2. Difference in shear strains between PDL and glue load cases at each gage location. The effect of the PDL decreases with distance from point of load application.

Table 6-6. Effect of loading rate on strain at Gage 1.

Specimen	Rate (mm/s)	Slope	Intercept	Difference in Slopes (P)
	0.2	0.396	45.67	
1	0.4	0.416	41.96	5.4×10^{-11}
	0.8	0.369	53.11	
	1.6	0.421	33.93	
	3.2	0.385	44.71	
	0.2	0.323	30.76	
2	0.4	0.347	45.63	2.8×10^{-33}
	0.8	0.357	54.38	
	1.6	0.374	63.34	
	3.2	0.385	73.22	
	0.2	0.421	17.67	
3	0.4	0.304	35.11	1.7×10^{-33}
	0.8	0.317	28.48	
	1.6	0.329	22.38	
	3.2	0.232	45.5	
	0.2	0.388	85.13	
4	0.4	0.436	62.25	1.7×10^{-24}
	0.8	0.435	48.94	
	1.6	0.417	23.98	
	3.2	0.379	63.63	
	0.2	0.92	77.72	
5	0.4	0.917	85.67	2.5×10^{-17}
	0.8	0.913	92.74	
	1.6	0.919	92.84	
	3.2	0.916	94.55	

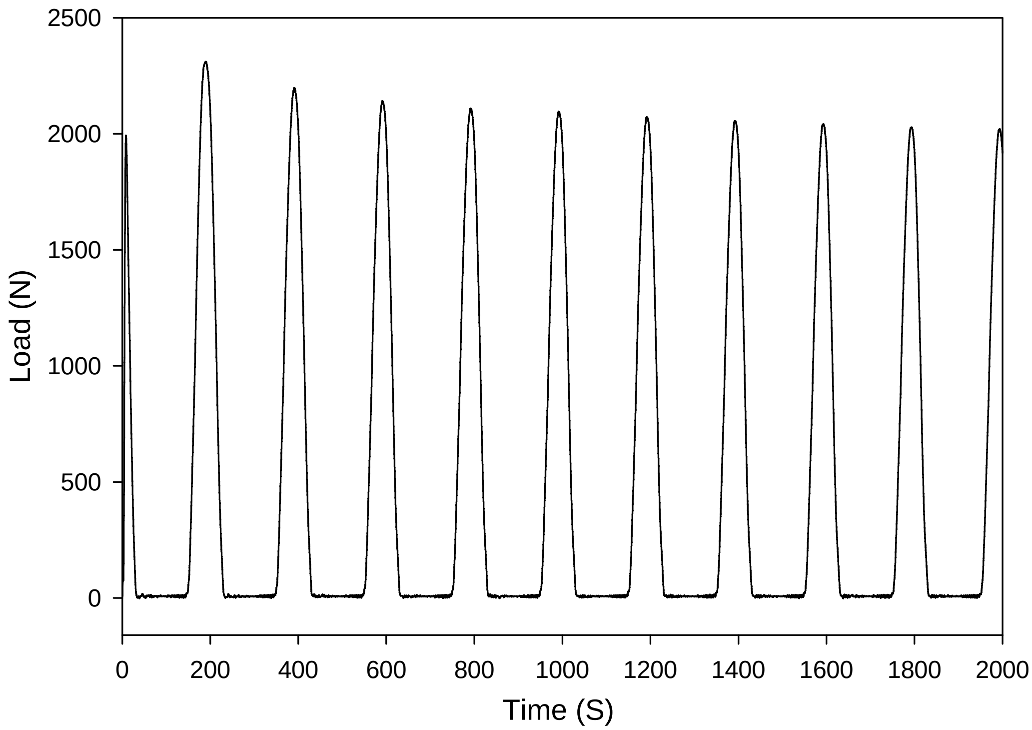

Figure 6-3. Load recovery during cyclical loading for intact PDL. Demonstrates that load removal is immediate with no lag time.

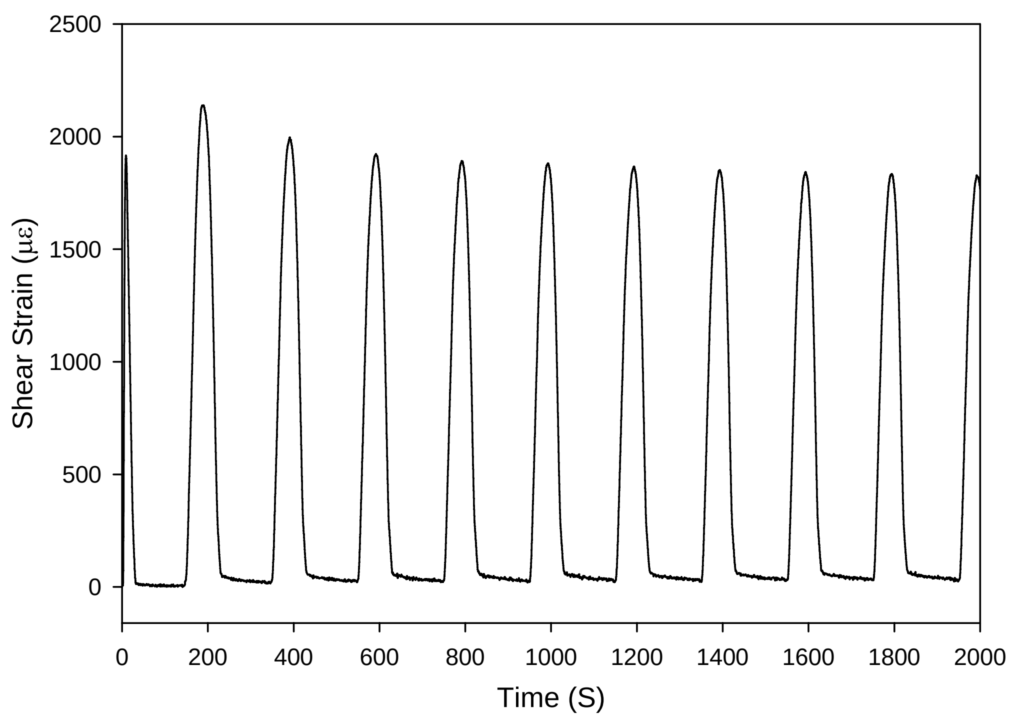

Figure 6-4. Shear strain recovery during cyclical loading for intact PDL. Although load removal is immediate strain recovery demonstrates hysteresis.

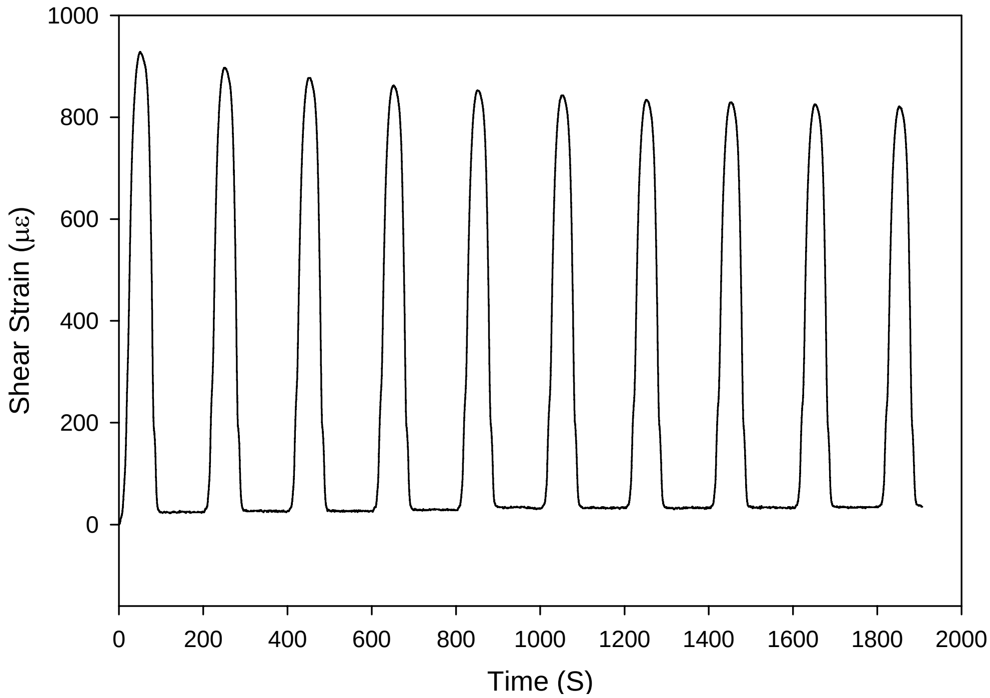

Figure 6-5. Shear strain recovery during cyclical loading for glued tooth. Unlike PDL intact tooth, bonded tooth does not demonstrate similar levels of hysteresis.

Table 6-7. Shear strain at end and start of loading cycles for all frequencies[1].

Frequency (Hz)	Cycles	Shear (με) at Load Cessation		Shear (με) at Load Application	
		PDL	Glue	PDL	Glue
0.5	1-3	37 ± 20	26 ± 0.5	16 ± 12	25 ± 1.5
	52-54	55 ± 1	55 ± 1	40 ± 4	53 ± 1
1	1-3	27 ± 15	67 ± 4	19 ± 6	50 ± 6
	98-100	38 ± 7	70 ± 3	30 ± 0.6	58 ± 0
2	1-3	11± 8	28 ± 19	8 ± 5	18 ± 11
	125-127	30 ± 5	57 ± 3	27 ± 3	51 ± 2

[1] Strains reported represent the values offset from zero.

CHAPTER 7
DISCUSSION

PDL function during occlusal loading is an important variable in a clear understanding the biomechanics of the jaws and face; however, its influence on mechanical behavior of alveolar bone is poorly understood. These experiments are an effort to understand the role of the PDL in distribution of bite forces and to provide experimental evidence to inform models of the mechanical environment of the jaws and face. The interpretive consequences of simplifying or ignoring the PDL in finite element models are not well understood. The experimental data presented here achieves multiple aims: 1) to compare the strain response of a cranium to a load applied through an intact PDL versus a load applied through simulated direct tooth-bone contact, 2) to consider the effects of the PDL in strain dissipation local to and remote from the point of load application and 3) to establish the degree of rate dependent strain behavior observed in alveolar bone and 4) to evaluate the degree of hysteresis observed in alveolar bone in the two loading environments.

Hypothesis 1: Removal of PDL Has an Effect on Facial Strain

Maximum shear strains were markedly different after the tooth was bonded using cyanoacrylate adhesive. Since the PDL mediates force transfer from tooth to bone, this is not surprising. One specimen (2) did not demonstrate drastic changes in strain magnitude which may have been the result of an incompletely destroyed PDL. In this specimen tooth removal was not accomplished with the root structure intact. It is likely that parts of the multi-rooted tooth's PDL were not completely mechanically destroyed during this process, perhaps resulting in the similarities between the two experiments. It is assumed that the PDL acts as a shock absorber to load (Matsuo and Takahashi 2002) and therefore the strains were expected to be overall higher in the bonded tooth load scenario.

The PDL serves to reduce the stress in surrounding alveolar bone but the exact mechanism of this behavior is unclear. It was initially thought to act by cushioning the blow like a sealed container similar to the action of a balloon. However, more recently it has been suggested that it more closely resembles that action of "pushing liquid through a syringe", since the PDL is highly vascularized and not a closed system (Matsuo and Takahashi 2002). In normal occlusion the teeth are subjected to high loads, more than 700N in humans (Atkinson and Ralph 1976). If there were no load buffering the gracile morphology of the alveolar region would likely fail under normal use. A major concern in dental implants is mitigating stress concentrations in the alveolus (Carvalho et al. 2004).

The established load buffering behavior of the PDL indicates that its removal would likely produce higher strain in the alveolus. However, its shock absorber behavior has only been addressed in terms of an overall reduction in transmitted force; the PDL may also serve to redistribute loads along the alveolus. If the PDL does redistribute loads away from the delicate cervical region of the alveolus and instead towards the face, upon removal of the PDL a decrease in strains at remote locations is expected. Without the PDL in place to shunt stress to more remote locations strain is likely to increase cervically and but also may decrease at some locations remote to the alveolus.

In three out of four cases (including the minor response in Specimen 2) strains were predominantly higher in the glued case, excluding Gage 1 of Specimen 1. Similarly, the slope of the load-deformation curve at all gage locations was higher for the glued tooth in the same three cases, again excluding Gage 1 of Specimen 1. Since per unit load higher strains were observed, this leads to the conclusion that glue distributed a higher proportion of the load to the alveolar bone than the PDL in those three cases, illustrating the shock absorbing behavior of the PDL.

The reverse finding is observed for Specimens 4 and 5, the shear strains were higher for the intact PDL at all gage locations (except for Specimen 4, Gage 1 at the lowest load rate). Accordingly, those two specimens also indicated more effective load transfer via the PDL based on the slopes of the load deformation curve. It is possible that in these two cases the bonded tooth served to redistribute load away from the gage sites on the alveolar bone and towards the root tips and deeper cortical bone, since the strains were also higher at the remote gages. This is an incomplete explanation as the strain gradient was similar between the bonded and intact PDL experiments in those two specimens, only the maximum values of the strains differed. Gage placement may also have affected these results. Although care was taken to approximate gage placement in all specimens, exact reproduction of gage locations was impossible due to small differences in specimen morphology. While these strain values might be an artifact of the bonding process or of individual specimen morphology it seems more likely that they might be the result of angled load application. If the applied load was not purely occlusal resultant tipping of the tooth could produce unusual strain values (Ren et al. 2002).

Hypothesis 2: Strain Gradients Increase Apically and Distance Reduces PDL Effects

Based on the morphology of alveolar processes and the role of the PDL in load distribution it was predicted that the strains would be higher in the apical region for an intact PDL, similar to the results found by Asundi and Kishen (2000). There was a gradient found in both the PDL and glued experiments and in the majority of cases loads were largest at the remote rosette, supporting earlier findings. In two cases Gages 1-3 created a very clear gradient with Gage 1 reporting the smallest strains, Gage 3 the largest and Gage 2 exhibiting intermediate strains, which were exactly as anticipated. In one case the second gage produced the highest strains and in two others the second gage produced the lowest strains. So although the exact gradient in the cervical region of the alveolar process was somewhat unclear, strains were consistently larger in the apical direction.

The gradient established in the PDL experiments was well replicated after tooth bonding in two specimens; however the magnitude of strains composing the gradient tended to vary. This provides additional insight into the functional PDL-alveolar bone unit. Upon removal of the PDL the strain distribution pattern remains similar, but the gross values are affected. In these cases it appears that the PDL acted to buffer the alveolar region from overall strain magnitude and had little effect on the pattern of strain distribution. Specimen 4 only partially follows this trend and may be considered a third example of this behavior except for at the two lowest load rates. At the lowest load rate strains are reduced at Gages 1 and 3 but increase at Gage 2. At 0.4mm/s Gages 2 and 3 follow the PDL pattern but Gage 1 demonstrates an atypical drop in strains. The variation in this specimen may be an artifact of experimental variation such as angled load application, however a similar change in pattern is also observed for two other specimens.

In Specimen 1 the PDL apparently served to shunt stress cervically. At all load rates the strains recorded at Gage 3 remain similar between the two load cases, but for the destroyed PDL Gage 1 strains markedly decrease and at Gage 2 strains increase. In this specimen there is a clear reorganization of the pattern of strain distribution with the PDL serving to reduce strain at Gage 2 and apparently displace it cervically. Specimen 3 demonstrates a similar pattern. With the PDL intact Gage 1 demonstrates higher strains than Gage 2 while the reverse is seen after the PDL is destroyed. This pattern is consistent for all load rates except for at 3.2mm/s where an apically increasing strain pattern is seen for the both the intact and destroyed PDL. These two specimens indicate a more complex behavior for the PDL than simple reduction in transmitted loads. Based on these data the “shock absorber” behavior of the PDL is very complex and warrants further investigation.

The effect of the PDL in load buffering decreases as distance increases from the point of load application. There is a rather uniform decrease in difference of both strain magnitude and slope as distance increases, indicating that for regions of study that are far removed from the point of load application the effects of the PDL are likely negligible. However as the current gages locations under study demonstrated measurable differences in behavior between the two load cases it should be assumed that the mechanical response of the immediate alveolar region is still greatly affected by the presence of the PDL.

Hypothesis 3: Rate Dependent Strains Expected for the Intact PDL

The PDL and alveolar bone are both viscoelastic and rate dependent materials. Accordingly rate dependent strain levels were anticipated. This was observed in all specimens. It has been established for the PDL that at higher loading rates there is more effective force transfer from tooth to bone (Komatsu and Chiba 1993, Chiba and Komatsu 1993). Higher rates of loading would likely increase the stiffness of the PDL (Komatsu and Chiba 1993) supporting the initial conclusions that the bonded tooth would produce higher strains than an intact PDL. In the present experiments no clear pattern of rate dependency could be identified for an intact PDL. Variation in direction of bite force has a high impact on local strains (Hylander et al. 1998). It is probable that our inability to precisely control bite force direction between specimens precluded discovery of a discernible pattern of rate dependency in these experiments.

Hypothesis 4: Hysteresis Apparent for Intact PDL but Reduced in Bonded Tooth

Following our prediction, hysteresis was clearly observed in all intact PDL experiments. It was also assumed that since bone is viscoelastic, there would be some smaller level of hysteresis present in the bonded tooth runs. At 0.5Hz hysteresis was present in the intact PDL experiments but nonexistent for the bonded tooth experiments. At higher rates hysteresis was evident in the bonded tooth experiments, but to a lesser degree than the intact PDL. This reinforces the suggestion that the PDL's viscoelastic behavior can be seen in the alveolar bone response to loading. These observations support the inclusion of the PDL in regional models. Loading of the masticatory system is dynamic and therefore hysteresis is probably operative physiologically. Depending on the questions being modeled, these results indicate that the PDL and its dynamic behavior should be accounted for.

CHAPTER 8
CONCLUSIONS

The PDL plays important roles in tooth support, proprioception and regulation of alveolar bone remodeling. The PDL has also been established as the medium of occlusal force transfer in the jaws although the exact nature of transfer is unclear (McCulloch et al., 2000). Its response to load has been documented as non-linear, viscoelastic and rate dependent (Komatsu and Chiba, 1993; Chiba and Komatsu 1993; Pini et al., 2002; Toms et al., 2002; Wills and Picton, 1978). The PDL has also been suggested as a “shock absorber”, functioning to reduce the occlusal force applied to surrounding alveolar bone (Matsuo and Takahashi 2002) As the PDL is responsible for the distribution of bite force, it follows that these unique properties of the PDL have an effect on resultant strain dissipation in the skull. Whether this effect is an important consideration for theoretical modeling of the stress field in the skull is unclear.

These experiments are an effort to understand the role of the PDL in distribution of bite forces and to provide experimental evidence to inform models of the mechanical environment of the jaws and face. While some authors of finite element models of the skull have included the PDL in their analyses (Atmaram and Mohammed, 1981; Korioth et al., 1992) others have chosen to disregard the potential role of the ligament in force transfer (Knoell, 1977; Hart and Thongpreda, 1988; Hart et al, 1992; Marinescu et al., 2005; Richmond et al., 2005). The interpretive consequences of this choice are generally unknown, given our ignorance of local and remote effects of the PDL on strain distribution. Accordingly, I presented experimental data specifically designed to assess the effects of the PDL on facial strain magnitude and patterns resulting from occlusal loads.

Of primary interest is the load transmitting behavior of the PDL. It has been established as the mediator of load transmission but the exact behavior of force transfer is not clear. These data were anticipated to provide a more accurate understanding of the role of the PDL in transferring stress to alveolar bone. It is clear that the PDL does act in some basic “shock absorber” capacity, however, the exact behavior is more complex than pure hydraulic dampening of loads. Based on the results of this study, the PDL likely also serves to redirect loads to create a more uniform strain field around the alveolus although the exact advantage of the redistribution pattern observed here is somewhat unclear. The benefit of the PDL to the alveolus is likely two-fold, to serve as a damper of masticatory stress and also to redistribute stress in a manner that protects alveolar architecture.

The role of the PDL in creating strain gradients in the alveolus requires more study. It would be informative to compare the behavior observed in this study to that predicted by finite element analysis. Rees (2001) created a finite element model of a tooth and modeled its stress pattern after simplification or removal of the dental supporting structures. It would be useful to generate a similar model of alveolar bone and tease out the individual influences of the crown structure, root morphology and the PDL on alveolar stress distributions. A clear understanding of the integration of these structures in mediating masticatory forces is clinically important in maintaining alveolar health after dental reconstruction.

The behavior of the PDL also seems likely to change depending on where in the tooth row it is being analyzed; future work might involve the investigation of differential behavior of the PDL along the tooth row. Functional adaptations of the PDL to different loading regimes may indicate masticatory adaptation (Tanaka et al. 2006) and a clear understanding of how the role of the PDL in force distribution changes according to load history might be informative in comparative studies.

Finite element models are useful for examining the stress distribution of dental structures as the methods can cope with the complex geometry of the masticatory system. This study highlights the importance of including the PDL in finite element studies of the dentition and mastication. However, based on the observed strain gradients I infer that effects of the PDL on bone strain are primarily local to the load application site. Accordingly if the goal of comparative or mathematical models is to assess strain patterns local to the alveolar process, accounting for the mechanical properties of the PDL is essential. In whole skull models, or those with interests in stress patterns remote to the teeth, simplification of the PDL likely has minimal interpretive cost.

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