An In-Vitro Analysis of the Stresses in
Natural and Restored Human Teeth

Thesis for the degree of Doctor of Philosophy
in the Faculty of Medicine of the University of London
submitted by

NEIL MEREDITH BDS.,MSc.,(U.Lond.)

Department of Conservative Dentistry
Institute of Dental Surgery
London, WC1X 8LD.

May 1992
ABSTRACT

The distribution of stresses on intact, prepared and restored teeth during restoration and subsequent loading has been investigated in-vitro using a number of techniques.

A review of the relevant literature covering experimental and numerical stress analysis techniques and their applications is given together with an assessment of the physical and mechanical properties of teeth and aesthetic dental restorative materials. In-vivo measurements of occlusal loads and in-vitro loading techniques to determine the fracture resistance of prepared and restored teeth are also discussed.

A Knoop indentation technique was used to determine values for the micro-hardness and Young's modulus of human enamel and dentine. Results obtained were in good agreement with those published in the literature. Young's modulus and Poisson's ratio were established for a range of composite and polyalkenoate materials in four point bending and uniaxial tension, tested at a range of cross-head speeds and temperatures (0.05, 0.1, 0.2 mm/min at 23°C and 35°C).

Full-field stress pattern analysis of intact, prepared and restored teeth under dynamic loading by measurement of thermoelastic emission revealed a significant difference in the distribution of stresses between specimen types. Regions of high stress were identified as potential sites for more detailed analysis using strain gauges.

Rosette gauges mounted on teeth in-vitro were subjected to a range of restorative procedures and loaded (10-200N) at different rates (0.01-0.1 KNs\(^{-1}\)) and positions. Strains measured during the restorative and loading procedures were used to calculate the magnitude and direction of the principal strains. Cavity preparation resulted in an increase in strains which was not reduced by placement of an amalgam or bonded composite restoration.

Finite element models of the experimental specimens were used to investigate the stress distribution at the tooth-restoration interface, the effects of applied loads and the properties of restorative materials.
ACKNOWLEDGMENTS

I owe a debt of gratitude to many people for their discussions - too many to acknowledge them all properly here.

Firstly, I am greatly indebted to my supervisors, Mr. D.J. Setchell and Professor S.A.V. Swanson for their advice and encouragement throughout the period of this work. Secondly, I am very grateful to Dr. M. Sherriff for his discussions and advice on the statistical analysis of the data. The assistance he provided with the development of the computer program described in Appendix IV was invaluable.

Thirdly, I am indebted to the Medical Research Council for the award of a training fellowship (G84/2083) which enabled me to undertake this research and would like to add my thanks to the staff of the training awards group who were most helpful.

My thanks also go to Mrs. R. Randall and Dr. R. L. Erickson of 3M dental products for their discussions and providing the experimental materials used in a large part of this study.

In addition I am indebted to Mr. B. Barnett of Ometron Ltd. for kindly providing his time and loan of the SPATE instrumentation used in Chapter 4.

I would like to thank Dr. M. Punjani of the University of London Computer Centre for discussing applications of finite element analysis and enabling me to evaluate a range of programs.

Dr. A. Bell and Dr. A. Newton of FEA Ltd. patiently discussed the implementation of LUSAS, the finite element software package used in Chapter 6 and I am very grateful to them for this as well as reproducing some of the figures in colour.

I am also grateful to Mr. D. Murphy and Mr. C. Chaperlin of the department of Mechanical Engineering at Imperial College of Science and Technology for fabricating the specimen loading rig and performing numerous modifications to existing equipment so skillfully.

Special thanks must go to Barbara and to my parents for their encouragement, support and patience throughout this work.
'...The power of any spring is in the same proportion with the Tension thereof: That is, if one power stretch or bend it one space, two will bend it two, three will bend it three, and so forward. And this is the Rule or Law of Nature, upon which all manner of Restituent or Springing motion doth proceed...'

Robert Hooke, 1676
## CONTENTS

Title page 1  
Abstract 2  
Acknowledgements 3  
List of contents 5  
List of figures 11  
List of tables 18  
Notation 20

### CHAPTER 1 -  
LITERATURE REVIEW AND AIMS OF THE PRESENT STUDY 24

1.1 Introduction 25  
1.1.1 Limitations of established restorative materials 25  
1.1.2 Recent developments in restorative materials 26  
1.1.3 How far is structural restored in prepared teeth 27  
1.2 Objectives of the present work 28  
1.3 Experimental stress analysis techniques 30  
1.3.1 Introduction 30  
1.3.2 Full-field analysis methods 30  
1.3.2.1 Brittle coatings 30  
1.3.2.2 Photoelastic techniques 31  
1.3.2.3 Thermoelastic stress analysis techniques (SPATE) 33  
1.3.3 Discrete field analysis methods 37  
1.3.3.1 Strain gauges and applications 37  
1.3.3.2 Use of foil resistance gauges for strain measurement in dental research. 38  
1.3.3.3 Summary of experimental stress analysis methods and their application in dental research 41  
1.4 Numerical stress analysis techniques 42  
1.4.1 Introduction 42
1.4.2 Application of the finite element method in dental research

1.4.2.1 Model geometry

1.4.2.2 Material properties

1.4.2.3 Boundary conditions and interfaces

1.4.2.4 Load Cases

1.4.2.5 Validation

1.4.2.6 Summary of the applications of finite element analysis in the measurement of the stress distribution in sound and restored teeth

1.4.3 Failure in structures

1.4.3.1 Yield and fracture criteria

1.5 The Physical properties of teeth

1.5.1 Introduction

1.5.2 Determination of the mechanical properties of teeth using static test methods

1.5.3 Determination of the mechanical properties of teeth using dynamic test methods

1.5.4 Hardness and fracture characteristics of enamel and dentine

1.5.5 Summary of the physical properties of teeth

1.6 The Mechanical properties of aesthetic restorative materials

1.6.1 Introduction

1.6.2 Mechanical testing methods for restorative materials

1.6.3 The influence of material composition and the degree of polymer conversion on the mechanical properties of restorative composites.

1.6.4 The influence of environmental conditions on the mechanical properties of composite materials

1.6.5 Summary of the mechanical properties of composite materials

1.7 In-vivo and In-vitro loading of teeth

1.7.1 Introduction
1.7.2 In-vivo measurement of occlusal forces 74
1.7.3 In-vitro simulation of functional occlusal loading 78
1.7.4 In-vitro measurement of the fracture resistance of natural and restored teeth
   1.7.4.1 Load application 80
   1.7.4.2 Specimen selection and mounting methods 82
   1.7.4.3 Restorative Procedures 86
   1.7.4.4 A comparison of in-vitro test methods with the clinical performance of restored teeth 89
1.7.5 Summary of in-vitro and in-vivo methods for determining the fracture resistance of sound and restored teeth 90

CHAPTER 2 -
MEASUREMENT OF MICRO-HARDNESS AND YOUNG'S MODULUS OF HUMAN ENAMEL AND DENTINE

2.1 Aims and objectives 92
2.2 Method 93
   2.2.1 Specimen preparation 93
   2.2.2 Instrumentation 94
   2.2.3 Test procedure 94
2.3 Results 96
2.4 Discussion 101
   2.4.1 Discussion of the method 101
   2.4.2 Discussion of the results 104

CHAPTER 3 -
MEASUREMENT OF THE MECHANICAL PROPERTIES OF AESTHETIC RESTORATIVE MATERIALS

3.1 Aims and objectives 110
3.2 Method 111
   3.2.1 Specimen materials 111
CHAPTER 5 -
IN-VITRO DISCRETE FIELD STRESS ANALYSIS OF INTACT AND RESTORED TEETH AND MODEL SYSTEMS USING STRAIN GAUGES.

5.1 Aims and objectives 161

5.2 Method 163

5.2.1 Specimen preparation - natural teeth 163
5.2.2 Specimen fabrication - composite model system 165
5.2.3 Cavity preparation - natural teeth 167
5.2.4 Cavity restoration 167
5.2.5 Instrumentation 171

5.2.5.1 Load application and control 171
5.2.5.2 Loading rig - design and fabrication 173
5.2.5.3 Loading rig - compliance 175
5.2.5.4 Strain measurement 175
5.2.5.5 Strain gauge instrumentation 177
5.2.5.6 Data acquisition 177

5.2.6 Test Procedure 179

5.3 Results 183

5.3.1 Calibration 183
5.3.2 Restorative procedures 190
5.3.3 Loading procedures 190

5.4 Discussion 205

5.4.1 Discussion of the method 205
5.4.2 Discussion of the results 206

5.4.2.1 Restorative procedures 206
5.4.2.2 Loading procedures 209
CHAPTER 6 -
IN-VITRO STRESS ANALYSIS OF PREPARED AND RESTORED HUMAN TEETH USING THE FINITE ELEMENT METHOD

6.1 Aims and objectives
6.2 Method
   6.2.1 Specimen preparation
   6.2.2 Pre-processing
   6.2.3 Analysis
6.3 Results
6.4 Discussion of the method and results

CHAPTER 7 -
CLINICAL IMPLICATIONS, PROPOSED FURTHER WORK AND CONCLUSIONS.

7.1 Clinical implications
   7.1.1 The effects of cavity preparation on teeth subjected to simulated functional loads
   7.1.2 The effects of cavity restoration on cuspal strains
   7.1.3 Cuspal stresses and strains on restored teeth in-vitro
7.2 Proposed further work
7.3 Conclusions

Appendix I - Application of strain gauges to biological structures
Appendix II - The thermal behaviour of foil resistance strain gauges
Appendix III - Strain gauge amplifier design
Appendix IV - The derivation of principal strains and angles from experimental strain gauge measurements
Appendix V - Materials
Appendix VI - The influence of the material properties of embedding media in in-vitro load testing

References
LIST OF FIGURES

Figure 1. Programme of work 29

Figure 2. SPATE instrumentation for thermoelastic stress analysis
showing scanner head (A) and equipment rack (B) 35

Figure 3. Representative occlusal loading curves taken from three studies 77

Figure 4. Schematic of servo-hydraulic testing machine to simulate
mastication (after Delong and Douglas, 1991) 79

Figure 5. (a) Location of occlusal contacts on a lower first permanent molar 83
(b) Excursive pathways of opposing cusps on a lower first permanent
molar tooth (after Shillingburg, Wilson & Morrison, 1984) 85

Figure 6. Potential sites of occlusal interferences between upper and lower
premolar teeth 85

Figure 7. (a) Mean Knoop hardness in enamel with distance from the tooth surface 98
(b) Mean Knoop hardness in dentine with distance from the amelodentinal junction 98

Figure 8. (a) Mean Young’s modulus in enamel with distance from the tooth surface 100
(b) Mean Young’s modulus in dentine with distance from the amelodentinal junction 100

Figure 9. Scanning electron micrographs of Knoop indentations in
(a) enamel and (b) dentine surfaces 102

Figure 10. (a) Variation of residual Knoop impression dimensions with hardness/modulus ratio for enamel 107
(b) Variation of residual Knoop impression dimensions with hardness/modulus ratio for dentine 107

Figure 11. Four point bending rig for aesthetic restorative materials 114

Figure 12. Schematic diagram of instrumentation used for flexural testing of restorative materials in four point bending 116
<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>13</td>
<td>Schematic diagram of instrumentation used for tensile testing of restorative materials</td>
</tr>
<tr>
<td>14</td>
<td>Load and displacement curves plotted against time for a composite specimen loaded in four point bending</td>
</tr>
<tr>
<td>15</td>
<td>Variation in displacement with load for composite materials subjected to four point bending at different cross-head speeds</td>
</tr>
<tr>
<td>16</td>
<td>Relevant dimensions for the calculation of the deflection of a beam in four point bending</td>
</tr>
<tr>
<td>17</td>
<td>Axial and tensile strain plotted against load for a range of materials at 23°C and a crosshead speed of 0.1mm/min</td>
</tr>
<tr>
<td>18</td>
<td>Axial vs. transverse strain for a range of materials loaded in tension at a cross-head speed of 0.1mm/min at 23°C</td>
</tr>
<tr>
<td>19</td>
<td>Finite element mesh and contour plot of maximum principal stresses on a tensile composite specimen with bonded strain gauges attached</td>
</tr>
<tr>
<td>20</td>
<td>Strains measured on the surface of a tensile composite specimen having a range of values of Young's modulus and subjected to a load of 25N - derived using the finite element method</td>
</tr>
<tr>
<td>21</td>
<td>Variation in peak axial strain with time for a polyalkenoate specimen loaded in tension</td>
</tr>
<tr>
<td>22</td>
<td>Schematic diagram of instrumentation used for thermoelastic stress analysis of teeth</td>
</tr>
<tr>
<td>23</td>
<td>Range of the principal stresses observed on the surface of teeth subjected to a range of operative procedures by measurement of thermoelastic emission</td>
</tr>
<tr>
<td>24(a)</td>
<td>SPATE plot of the principal stresses on the buccal surface of an unprepared tooth subjected to a dynamic load of 200N</td>
</tr>
<tr>
<td></td>
<td>(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of an unprepared tooth</td>
</tr>
<tr>
<td>25(a)</td>
<td>SPATE plot of the principal stresses on the buccal surface of a</td>
</tr>
</tbody>
</table>
prepared tooth subjected to a dynamic load of 200N

(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of a prepared tooth

Figure 26. (a) SPATE plot of the principal stresses on the lingual surface of a prepared tooth subjected to a dynamic load of 200N

(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of a prepared tooth

Figure 27. (a) SPATE plot of the principal stresses on the buccal surface of a tooth restored with a bonded composite material and subjected to a dynamic load of 200N

(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of a tooth restored with a bonded composite material

Figure 28. Dimensions of geometric composite models

Figure 29. Rosette strain gauge mounted on the buccal surface of a lower first molar and indicating the level of the cavity floor.

Figure 30. Relationship between buccal and lingual strain gauge and orientation relative to the occlusal plane.

Figure 31. In-vitro loading positions on natural teeth and composite specimens

Figure 32. Timing diagram illustrating rate and magnitude of load application to specimens

Figure 33. 3rd angle projection of specimen loading rig

Figure 34. Photograph of specimen mounting assembly and method of load application (Load position 1)

Figure 35. Schematic for specimen loading and strain measurement instrumentation

Figure 36. Sequence of load application, restorative procedures and strain measurement

Figure 37. Load-displacement plots for different specimen types subjected to...
axial and angular loads

Figure 38. Frequency response of rosette type strain gauge mounted on a composite specimen and subjected to 100N load at different rates

Figure 39. Apparent strain in relation to time and temperature over periods of twenty minutes and twelve hours

Figure 40. Derivation of principal and shear strains in a specimen prepared with an MOD cavity and subjected to a range of intercuspal loads at different rates

Figure 41. Graphical representation of the magnitude and direction of the principal and shear strains on the buccal and lingual surfaces of an intact tooth subjected to an intercuspal load of 100 Newtons

Figure 42. Changes in strain during placement and polymerisation of a bonded composite restoration in a molar tooth

Figure 43.(a) Mean cumulative principal and shear strains during placement of MOD bonded composite restorations in composite models
(b) Mean cumulative principal and shear strains during placement of occlusal bonded composite restorations in natural teeth
(c) Mean cumulative principal and shear strains during placement of MOD bonded composite restorations in natural teeth

Figure 44. Mean strains for teeth prepared with MOD cavities and subjected to intercuspal loads

Figure 45. Variation in principal and shear strains with load and position in a natural tooth prepared with an MOD cavity

Figure 46. Mean principal and shear strains and angles for a geometric composite molar model subjected to a load of 100 Newtons in a range of positions

Figure 47. Mean principal and shear stresses on teeth prepared with MOD cavities and subjected to an intercuspal load, calculated using a range of values for Young's modulus

Figure 48. Mean strains on intact and restored teeth loaded intercuspally to
100 Newtons

Figure 49. (a) Mean principal strains on intact natural teeth loaded to 100 Newtons in two positions
(b) Mean principal strains on teeth restored with bonded composite restorations loaded to 100 Newtons in two positions
(c) Mean principal strains on natural teeth with MOD preparations loaded to 100 Newtons in two positions
(d) Mean principal strains on natural teeth restored with MOD amalgam restorations loaded to 100 Newtons in two positions

Figure 50. Mean principal and shear strains on natural teeth subjected to a different operative procedures and loaded to 100 Newtons in a range of positions

Figure 51. Schematic representation of the section taken from a restored tooth used to provide the geometry for a finite element model

Figure 52. (a) Two dimensional plane-strain finite element mesh of a bucco-lingual section of a lower first molar tooth
(b) Material property assignment to finite element mesh
(c) Loading and measurement points on finite element mesh

Figure 53. Maximum and minimum principal stresses in a buccolingual section of a lower molar tooth restored with a bonded composite restoration and subjected to an intercuspal load of 200 Newtons - derived using the finite element method

Figure 54. (a) Distribution of the maximum principal stresses along section line 1 along the buccal tooth-restoration interface in the Figure (53)
(b) Distribution of the minimum principal stresses along section line 1 along the buccal tooth-restoration interface in Figure (53)
(c) Distribution of the maximum principal stresses along section line 2 along the lingual tooth-restoration interface in Figure (53)
(d) Distribution of the minimum principal stresses along section line 2 along the lingual tooth-restoration interface in Figure (53)
(e) Distribution of the maximum principal stresses along section line 3 taken through the strain gauges in Figure (53)

(f) Distribution of the minimum principal stresses along section line 3 taken through the strain gauges in Figure (53)

Figure 55. Principal stresses in a molar tooth prepared with an MOD cavity and restored with an adhesive composite material, subjected to an intercuspal load of 10-200 Newtons - derived using the finite element method

Figure 56. Mean - difference plot to compare experimental data derived from strain gauge measurements with results from a finite element analysis

Figure 57. Principal stresses in a buccolingual section of a molar tooth subjected to an intercuspal load of 100 Newtons and restored with an adhesive material of variable elastic modulus - derived using the finite element method

Figure 58. (a) Contour plot of the maximum and minimum principal stresses in a buccolingual section of a lower molar tooth restored with an adhesive material having a value for Young’s modulus of 5 GNm$^{-2}$ - derived using the finite element method

(b) Contour plot of the maximum and minimum principal stresses in a buccolingual section of a lower molar tooth restored with an adhesive material having a value for Young’s modulus of 50 GNm$^{-2}$ - derived using the finite element method

Figure 59. Principal stresses in a molar tooth prepared with an MOD cavity and restored with an adhesive composite material, subjected to a load of 100 Newtons in a range of positions - derived using the finite element method

Figure 60. Principal stresses at nodes on the buccal and lingual surfaces of a 2-D section of a lower molar loaded at 100 Newtons in different positions and prepared with cavities varying in size and restored
with a composite material
- derived using the finite element method

Figure 61. Relationship between the maximum principal stresses from an intercuspal load of 100 Newtons and the ratio of values for Young's modulus for enamel and dentine

Figure II.1 Position of thermocouples and strain gauges on test specimen

Figure II.2 Surface temperature rise on strain gauge with increasing power density (single gauge)

Figure II.3 Temperature rise inside specimen with increasing power density (single gauge)

Figure II.4 Surface temperature rise on strain gauge with increasing power density (dual gauge)

Figure II.5 Temperature rise inside specimen with increasing power density (dual gauge)

Figure III.1 Schematic diagram of strain gauge amplifier and unity gain buffer (single channel)

Figure VI.1 Finite element mesh, loading diagram and surface allocation for a specimen embedded in a mounting block
LIST OF TABLES

Table 1. Summary of in-vivo methods used for the determination of occlusal forces 75
Table 2. Summary of in-vitro loading studies of intact, prepared and restored teeth 81
Table 3. Variation in mean Knoop hardness in dentine with distance from the amelo-dentinal junction 97
Table 4. Variation in mean Knoop hardness in enamel with distance from the tooth surface 97
Table 5. Variation in mean Young's modulus in dentine with distance from the amelo-dentinal junction 99
Table 6. Variation in mean Young's modulus in enamel with distance from the tooth surface 99
Table 7. Mean values for Young's modulus for aesthetic restorative materials loaded in four point bending 126
Table 8. Mean values for Young's modulus for aesthetic restorative materials loaded in tension at a range of temperatures and cross-head speeds 128
Table 9. Poisson's ratio for aesthetic restorative materials loaded in tension 128
Table 10. The range of the sum of the principal stresses measured using the thermoelastic method on the surface of intact and restored teeth subjected to an occlusal load of 200N 151
Table 11. Mean changes in principal and shear strains during placement of occlusal and MOD type composite restorations in natural teeth 189
Table 12. Total strains produced during placement of direct composite, indirect composite and glass ionomer lined composite restorations in natural teeth 191
Table 13. Statistical analysis comparing magnitude of load and positions on 196
intact teeth and occlusal and MOD preparations

**Table 14.** Variations in the principal stresses on the buccal and lingual surfaces of an intact tooth subjected to an intercuspal load of 100N with a range of properties for enamel and dentine

**Table 15.** Validation of finite element analysis by comparison with experimental data

**Table 16.** A comparison of the stresses resulting from an intercuspal load of 200N in teeth restored with composite and glass ionomer lined composite restorations - derived using the finite element method
NOTATION

In general, terms have been defined where they first appear in the text and again as a point of clarification where they appear in a new chapter. Nevertheless, the following list may be helpful for reference purposes. Throughout the text SI units have been employed except where specifically stated to the contrary.

\[
\begin{align*}
A_o & \quad \text{original cross sectional area} \\
a & \quad \text{displacement vector} \\
a & \quad \text{length of short Knoop diagonal in the fully loaded state} \\
a' & \quad \text{altered length of short Knoop diagonal} \\
B_1 & \quad \text{strain from buccal strain gauge, element 1} \\
B_2 & \quad \text{strain from buccal strain gauge, element 2} \\
B_3 & \quad \text{strain from buccal strain gauge, element 3} \\
B_{\varepsilon_1} & \quad \text{maximum principal strain from buccal strain gauge} \\
B_{\varepsilon_2} & \quad \text{minimum principal strain from buccal strain gauge} \\
B_r & \quad \text{shear strain from buccal strain gauge} \\
B_{\theta_1}, B_{\theta_2}, B_{\theta_3} & \quad \text{the angle of the principal and shear strains } B_{\varepsilon_1}, B_{\varepsilon_2} \text{ and } B_r \text{ with respect to the angle of element [1] on the buccal strain gauge} \\
B_{\sigma_1} & \quad \text{maximum principal stress from buccal strain gauge} \\
B_{\sigma_2} & \quad \text{minimum principal stress from buccal strain gauge} \\
b & \quad \text{length of long Knoop diagonal in the fully loaded state} \\
b' & \quad \text{altered length of long Knoop diagonal} \\
C & \quad \text{specific heat} \\
C_p & \quad \text{specific heat at constant pressure} \\
C_e & \quad \text{effective specific heat} \\
c & \quad \text{wave velocity} \\
D & \quad \text{specimen diameter (in four point bending)} \\
D_{1000} & \quad \text{digital strain reading during } 1000\mu e \text{ calibration pulse} \\
D_0 & \quad \text{digital value at zero strain}
\end{align*}
\]
<table>
<thead>
<tr>
<th>Symbol</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>$D_M$</td>
<td>digital value at measured strain</td>
</tr>
<tr>
<td>$d$</td>
<td>depth of a Vickers indent</td>
</tr>
<tr>
<td>$d_1$</td>
<td>displacement at one of the loading points (in four point bending)</td>
</tr>
<tr>
<td>$E$, $E_x$, $E_y$, $E_z$</td>
<td>Young's modulus (in respective directions)</td>
</tr>
<tr>
<td>$E_p$</td>
<td>Young's modulus of a photoelastic material</td>
</tr>
<tr>
<td>$F$</td>
<td>force</td>
</tr>
<tr>
<td>$G$</td>
<td>shear modulus</td>
</tr>
<tr>
<td>$H$</td>
<td>hardness</td>
</tr>
<tr>
<td>$HK$</td>
<td>Knoop hardness number</td>
</tr>
<tr>
<td>$h$</td>
<td>specimen height (in four point bending)</td>
</tr>
<tr>
<td>$l$</td>
<td>second moment of area</td>
</tr>
<tr>
<td>$j$</td>
<td>specimen breadth (in four point bending)</td>
</tr>
<tr>
<td>$K$</td>
<td>stiffness matrix</td>
</tr>
<tr>
<td>$K$</td>
<td>bulk modulus</td>
</tr>
<tr>
<td>$K_t$</td>
<td>stress concentration factors</td>
</tr>
<tr>
<td>$K_m$</td>
<td>thermoelastic constant</td>
</tr>
<tr>
<td>$K_{IC}$</td>
<td>critical stress intensity factor</td>
</tr>
<tr>
<td>$L$</td>
<td>length</td>
</tr>
<tr>
<td>$L_o$</td>
<td>original length</td>
</tr>
<tr>
<td>$L_1$</td>
<td>strain from lingual strain gauge, element 1</td>
</tr>
<tr>
<td>$L_2$</td>
<td>strain from lingual strain gauge, element 2</td>
</tr>
<tr>
<td>$L_3$</td>
<td>strain from lingual strain gauge, element 3</td>
</tr>
<tr>
<td>$L_{\epsilon_1}$</td>
<td>maximum principal strain from lingual strain gauge</td>
</tr>
<tr>
<td>$L_{\epsilon_2}$</td>
<td>minimum principal strain from lingual strain gauge</td>
</tr>
<tr>
<td>$L_\sigma$</td>
<td>shear strain from lingual strain gauge</td>
</tr>
<tr>
<td>$L_{\epsilon_1}$, $L_{\epsilon_2}$, $L_{\sigma_2}$</td>
<td>the angle of the principal and shear strains $L_{\epsilon_1}$, $L_{\epsilon_2}$ and $L_\sigma$ with respect to the angle of element [1] on the lingual strain gauge</td>
</tr>
<tr>
<td>$L_{\sigma_1}$</td>
<td>maximum principal stress from lingual strain gauge</td>
</tr>
<tr>
<td>$L_{\sigma_2}$</td>
<td>minimum principal stress from lingual strain gauge</td>
</tr>
<tr>
<td>$l$</td>
<td>distance between the outer loading points (in four point bending)</td>
</tr>
</tbody>
</table>
M modulus of rigidity
N Newton
P axial load
density
q contact diameter of a Vickers indent
R force vector
R resistance
RD relative deformation (Morin, DeLong and Douglas, 1984)
RS relative stiffness (Morin, DeLong and Douglas, 1984)
S sum of the principal stresses
S\textsubscript{a} strain sensitivity of an alloy
S\textsubscript{g} gauge factor
SR stiffness ratio (Morin, DeLong and Douglas, 1984)
s semiangle between opposite pyramidal edges in a Vickers
indentation
T Absolute temperature
T\textsubscript{g} glass transition temperature
u distance between the inner and outer loading points (in four point
bending)
v, v\textsubscript{xy}, v\textsubscript{yx} Poisson's ratio (in respective directions)
W total force
W\textsubscript{f} work of fracture
x, y, z Cartesian coordinates
y constant (0.91, Lawn & Howes, 1981)
Z specific acoustic impedance
\alpha coefficient of linear expansion
\alpha\textsubscript{l} constant of proportionality
\alpha\textsubscript{T} coefficient of thermal expansion
\epsilon\textsubscript{0} strain at no load
\sigma\textsubscript{0} stress at no load
\( \varepsilon \)  
strain

\( \varepsilon_s \)  
absolute strain

\( \varepsilon_t \)  
threshold strain required to crack brittle coating

\( \varepsilon_1 \)  
maximum principal strain

\( \varepsilon_2 \)  
minimum principal strain

\( \varepsilon_3 \)  
principal shear strain

\( \Theta \)  
direction of the maximum principal stress or strain with respect to the x axis

\( \mu \varepsilon \)  
microstrain

\( \sigma_1 \)  
maximum principal stress

\( \sigma_2 \)  
minimum principal stress

\( \sigma_3 \)  
principal shear stress
CHAPTER 1

LITERATURE REVIEW AND AIMS OF THE PRESENT STUDY

1.1 Introduction

1.2 Objectives of the Present Work

1.3 Experimental Stress Analysis Techniques

1.4 Numerical Stress Analysis Techniques

1.5 The Physical Properties of Teeth

1.6 The Mechanical Properties of Aesthetic Restorative Materials

1.7 In-vivo and In-vitro loading of teeth
1.1 Introduction

Caries is a disease process in which acids fermented from dietary sugars by bacteria cause the destruction of dental hard tissue. This disease can be arrested and may even be reversible in its early stages by oral hygiene and use of fluorides. Carious lesions occur on the occlusal, interproximal and cervical surfaces of teeth. Once a lesion has penetrated the enamel and reached the dentine progress can be rapid and result in the widespread destruction of tissue. Caries often spreads laterally along the amelodentinal junction undermining the enamel. Treatment of a carious lesion is desirable to restore the structural integrity of the tooth and prevent a loss of pulpal vitality. Patients desire an aesthetic and functional dentition and this has led to the development of restorative materials suitable for this purpose. Treatment of carious lesions has traditionally been carried out using a rotary cutting instrument to gain access to and remove the diseased tissue and create a cavity form suitable for the retention of a restorative material.

1.1.1 Limitations of established restorative materials

The role of a dental restorative material is to preserve the aesthetics, form and function of a tooth. For many years these aims have been achieved in part by the use of plastic materials such as amalgam (a silver-tin-mercury alloy). Continued research and development has resulted in amalgam alloys that have a relatively high strength, good corrosion resistance and are reasonably tolerant of placement technique. However a disadvantage is unquestionably, its poor aesthetics and in addition concern has been expressed regarding potential toxicity from the presence of unbound mercury. Amalgam does not adhere chemically to enamel or dentine and retention of a restoration within a cavity is enhanced by the presence of mechanical undercuts, slots, grooves and pins. Castings, often of precious metal alloys have also been widely used to restore teeth.

Clinical failure of a restored tooth may occur as a result of poor restoration design or placement technique, secondary caries, environmental breakdown of the restorative material, or fracture. It has been shown in-vitro (Vale, 1959) that the presence of a cavity within a tooth may decrease its resistance to fracture. In an intra-coronal amalgam restoration there is no adhesion between the restoration and remaining tooth structure and
the tooth/restoration interface has negligible tensile strength. Therefore placement of such a restoration will not increase the structural strength of a prepared tooth. Cast restorations are commonly used in an attempt to strengthen and support the remaining tooth structure following preparation. There is usually negligible tensile strength between the cement lute and remaining tooth structure or restoration and castings are often designed as crowns or overlays so that the cement lute is not loaded in tension during function. Fracture of an intact or restored tooth occurs as the result of a single impact load or by fatigue due to repeated loading. Much work has been carried out in-vitro to investigate the fracture behaviour of sound and restored teeth subjected to a range of loads. In-vitro simulation of fatigue induced failure in teeth is difficult to carry out however, as in-vivo loading parameters have not been clearly defined.

1.1.2 Recent developments in restorative materials

In response to the demand for more aesthetic restorations, a number of tooth-coloured materials based on filled resins or polyalkenoate cements have recently become available for clinical use. These may be placed and contoured in a plastic state directly in the cavity and allowed to set or polymerised using a visible light source. Alternatively an 'indirect' technique may be employed in which the restoration is fabricated and polymerised on a model of the preparation out of the patients mouth. The restoration is cemented in place using a resin based luting cement. It has been suggested (Wendt, 1987) that advantages of an indirect technique may include a reduction in polymerisation shrinkage during placement, better aesthetics and an improvement in mechanical properties attributable to a higher degree of polymer conversion achieved by use of elevated temperatures (Dionysopolous & Watts, 1989).

In addition to their aesthetic properties these materials differ from amalgam in their ability to adhere to the remaining tooth structure. Polyalkenoate cements are considered to adhere to enamel and dentine directly. However in order to achieve adhesion at the composite/tooth interface enamel acid etching and use of a dentine bonding agent is necessary.
1.1.3 How far is structural strength restored in prepared teeth?

It has been suggested by a number of workers that adhesion between the tooth and restoration may strengthen the remaining tooth structure (Douglas, 1983; McCullock & Smith, 1986). In order to establish whether a restoration can restore the original strength of a tooth it is important to understand how teeth may fail.

Loads applied to teeth in-vivo may be divided, broadly, into three types;

i) Impact loads, in which a single tooth or number of teeth are subjected to a load at a high stress rate which may result directly in tooth fracture.

ii) Functional loads, where one or more opposing teeth contact in excursions of the mandible that take place during normal functional activities (mastication). Loads are usually of low magnitude (10-20N) but may result in tooth wear or fracture of teeth or restorations due to fatigue.

iii) Loads of high magnitude (> 100N) acting on few teeth may occur as a result of occlusal interferences and in grinding habits. This may result in wear, tooth or restoration fracture, failure of a cement lute, tooth mobility or pain.

In-vitro measurement of fracture resistance has commonly been performed to investigate the behaviour of restored teeth under loading but an alternative approach is to determine the stress distribution in intact and restored teeth under simulated clinical loads. This has been attempted in a small number of studies in-vitro (see Morin et al., 1988[a]) but with the following limitations;

a) Strain measurements have been carried out using only single element gauges placed in an arbitrary position on the tooth surface, recording strain over a small area and in only one direction.

b) Numerical stress analysis techniques, most commonly using the finite element method, have often been simplified using crude geometry, inappropriate load cases and poorly selected material properties.

c) Some investigators have used techniques for load application which poorly represent the in-vivo environment.
Values for mechanical properties of enamel, dentine and restorative materials applied in previous investigations to calculate stresses from experimentally measured strains have typically been obtained from sources whose values may represent extremes of the range.

1.2 Objectives of the present work

The overall objective of this work was to develop a better understanding of the mechanical behaviour of intact and restored human teeth under a range of loading conditions, with special reference to intracoronal restorations and aesthetic restorative materials.

The specific objectives were;

a) Measurement of the elastic coefficients of enamel, dentine and the restorative materials used in the investigation to enable calculation of stresses from measured strains and enable the application of appropriate material properties to numerical analyses;

b) To measure strains on the surface of intact and restored teeth in-vitro under a range of loading conditions representative of those encountered clinically;

c) To apply numerical stress analysis techniques to supplement the data obtained from the experimental methods;

d) To develop criteria based on the results obtained to optimise the design of restorations and aid the development of materials to maximise the structural properties of restored teeth.
Full-field analysis techniques

Discrete-field analysis methods

Numerical stress analysis

In-vivo & in-vitro loading of teeth

Physical properties of teeth

Physical properties of restorative materials

Thermoelastic stress analysis of teeth (Chapter 4)

Strain gauge measurements on teeth (Chapter 5)

In-vitro loading of teeth (Chapter 5)

Indentation measurements on enamel & dentine (Chapter 2)

Numerical stress analysis of teeth (Chapter 6)

Tensile & Flexural testing of restorative materials (Chapter 3)

Clinical implications (Chapter 7)

Proposed further work

Conclusions

Review of the Literature

Method

Results & Discussion

Figure 1. Programme of work
1.3 Experimental Stress Analysis Techniques

1.3.1 Introduction

Values of stress at points on the surface of a structure are often calculated from experimentally measured strains. There are a number of experimental techniques that are used to measure strain on a specimen surface. These may be broadly divided into those that provide information on a large area of the specimen itself (full field techniques), a relative point on the specimen surface and those that employ a model of the structure. Such stress analysis techniques have been widely used in dentistry and medicine. The results sometimes contained errors because the chosen method was inappropriate for the application or the analysis had been incorrectly carried out.

The biomechanics of teeth are extremely complex and the difficulties of measuring strains associated with their small size, complex geometry, inadequately described mechanical properties and poorly determined loading conditions should not be underestimated.

Stresses in structures of simple geometry may be normal or shear and can be deduced by simple calculation. However, specimen geometry and loading conditions are often complex and the state of stress can only be resolved by use of an experimental or numerical stress analysis technique. An example of a numerical method is finite element analysis in which the structure is modelled as a finite number of variables, or elements, whose displacements and stresses are calculated by performing an algebraic solution of a set of equations (NAFEM; 1984). Often, an individual analysis technique cannot provide all the required information and two or more different methods may be used to build up a complete picture of the stress distribution in a specimen and provide validation. In order to be able to determine the technique most suitable for the analysis of sound and restored teeth it is useful to review experimental and theoretical stress analysis methods and discuss how they have been applied in dentistry.

1.3.2 Full-field Analysis Methods

1.3.2.1 Brittle coatings

A brittle coating applied directly to a test specimen can indicate the principal strains present on the surface, their directions and if the magnitude has exceeded a threshold
Figure 1. Programme of work
1.3 Experimental Stress Analysis Techniques

1.3.1 Introduction

Values of stress at points on the surface of a structure are often calculated from experimentally measured strains. There are a number of experimental techniques that are used to measure strain on a specimen surface. These may be broadly divided into those that provide information on a large area of the specimen itself (full field techniques), a relative point on the specimen surface and those that employ a model of the structure. Such stress analysis techniques have been widely used in dentistry and medicine. The results sometimes contained errors because the chosen method was inappropriate for the application or the analysis had been incorrectly carried out.

The biomechanics of teeth are extremely complex and the difficulties of measuring strains associated with their small size, complex geometry, inadequately described mechanical properties and poorly determined loading conditions should not be underestimated.

Stresses in structures of simple geometry may be normal or shear and can be deduced by simple calculation. However, specimen geometry and loading conditions are often complex and the state of stress can only be resolved by use of an experimental or numerical stress analysis technique. An example of a numerical method is finite element analysis in which the structure is modelled as a finite number of variables, or elements, whose displacements and stresses are calculated by performing an algebraic solution of a set of equations (NAFEM; 1984). Often, an individual analysis technique cannot provide all the required information and two or more different methods may be used to build up a complete picture of the stress distribution in a specimen and provide validation. In order to be able to determine the technique most suitable for the analysis of sound and restored teeth it is useful to review experimental and theoretical stress analysis methods and discuss how they have been applied in dentistry.

1.3.2 Full-field Analysis Methods

1.3.2.1 Brittle coatings

A brittle coating applied directly to a test specimen can indicate the principal strains present on the surface, their directions and if the magnitude has exceeded a threshold
value. As the coating is very thin strains occurring on the specimen are transmitted through without attenuation. A brittle coating cracks at a threshold tensile strain, which is often quite high, typically 500 \( \mu \varepsilon \). Detection of the presence of a crack may be difficult and can be assisted by use of a dye penetrating agent. Coatings (Stresscoat; Measurements Group, USA (1)) are commonly composed of resins in a solvent and may be conveniently sprayed on to a specimen, the optimum thickness being between 0.10-0.20 mm. A major disadvantage of these materials is the difficulty in using them to measure compressive strains as this entails pre-stressing the specimen prior to their application. The sensitivity of brittle coatings to variations in temperature and humidity is a further disadvantage.

The principal stress in a specimen perpendicular to the direction of a crack in a coating can be calculated as;

\[
\sigma_i = E \frac{\varepsilon_i}{\tau}
\]

where

- \( \sigma_i \) = principal stress in the specimen at the location of crack in the coating
- \( E_p \) = modulus of elasticity of specimen material
- \( \varepsilon_i \) = threshold strain required to crack coating when specimen is subjected to a uniaxial state of stress. (taken from Dally & Riley, 1985)

Possibly because of their difficult placement technique, brittle coatings appear to have been used only infrequently in dental research. Craig and Peyton (1965) performed an in-vitro investigation to study the patterns of surface strain on fixed bridgework using this technique. Bridges which had been cemented onto brass dies, were coated with a brittle lacquer and subjected to a range of static loads. The position and direction of cracks in the coating indicated areas of high strain on the crowns. However, the magnitude and duration of the loads applied in this investigation could not be considered to be representative of those arising in a clinical situation.

### 1.3.2.2 Photoelastic Techniques

Photoelastic stress analysis utilises the phenomenon of temporary double refraction observable in transparent non-crystalline solids. Some materials which are optically iso-
tropic when free of stress become anisotropic when they are stressed. Subjecting a model made from a photoelastic material to a mechanical load will produce a series of dark fringes known as isochromatics. A further set of fringes termed isoclinics appear where the direction of the principal stresses in the model coincide with the axes of the polarising filters used to view the model. These can be used to determine the direction of principal stresses at any point. Calibration of a photoelastic material is necessary to compensate for variations due to the batch, temperature and age of the material.

A photoelastic analysis may be two or three dimensional. In two dimensions a model of the test object is fabricated from a sheet of photoelastic material which is loaded and placed in a polariscope. The fringe order is observed and the magnitude and direction of the principal stresses can be calculated using a compensation method. It is possible to produce a three dimensional photoelastic model using a stress freezing technique. If a model which is fabricated from a polymer that exhibits diphasic behaviour is heated to a critical temperature and allowed to cool while an applied load is maintained the deformations produced will be locked into the model. The model can then be sliced and internal stresses examined.

Investigations using both two and three dimensional photo-elastic models have been used in dental research to study the stress distribution cavity preparation, design of crowns and bridgework and the stresses produced by posts and pins in teeth.

Early work by Mahler and Peyton (1955) used two dimensional photoelastic models of buccolingual sections of molar teeth that had been restored with amalgam and gold inlay type restorations and loaded axially in compression. Their results were only qualitative and suggested that placement of a restoration may result in increased shear stresses at the tooth/restoration interface. In a subsequent paper Mahler (1958) attempted to make quantitative measurements of the levels of stress in a model of a disto-occlusal amalgam restoration in a deciduous molar tooth.

A number of workers (Granath 1963, Craig et al. 1967[a], [b]) have used photoelastic methods to measure the stresses in prepared teeth and have shown that cavities having a rounded form exhibit lower levels of stress than those having margins with acute angles. Unfortunately most of the applications of photoelastic stress analysis in dental research
have been qualitative, indicating the presence and position of fringe orders without attempting to make quantitative measurements. Granath (1964[a]) commented on the limited clinical application of analyses of teeth which had used photoelastic models and had a simple, regular geometry.

Granath (1964[b]) attempted to measure the strains at the tooth/restoration interface using single element strain gauges mounted on the occlusal edge of two-dimensional photo-elastic models. The results suggested that marginal fracture of the restoration may occur even if its adaptation to the cavity wall is satisfactory. However, it is doubtful if comparisons could be drawn with a clinical situation because of significant differences in the mechanical properties between the photo-elastic resin used and human enamel.

Davidson (1964) was the first worker to produce a three-dimensional photoelastic model using the stress freezing technique. Johnson et al. (1968) utilised this method to produce models of a lower first molar prepared with occlusal cavities having sharp or rounded internal line angles. Results of a simple quantitative analysis showed that rounded internal line angles resulted in lower stresses. It was also reported that complete quantitative analysis of such a model would be an extremely complex task.

Photoelastic stress analysis is a sensitive and accurate technique, however its applications in restorative dentistry are limited by a number of factors;

a) **Material properties**: If a photoelastic model is composed of two materials representing enamel and dentine for example, it is important that ratios of the elastic modulus of the two materials are correct. The elastic modulus ratio for enamel:dentine is approximately 7:1 and this is difficult to replicate using photoelastic materials.

b) **Loading conditions**: Photoelastic models are loaded and viewed under static conditions which may not be representative of in-vivo loads.

c) **Model analysis**: Quantitative analysis of the stresses in photoelastic models of dental structures which have a complex geometry can be difficult.

### 1.3.2.3 Thermoelastic stress analysis techniques (SPATE)

Stress Pattern Analysis by Thermoelastic Emission (SPATE) is a full-field stress
analysis method based on the thermoelastic phenomenon described by Kelvin in 1853. The thermoelastic phenomenon occurs in solids and is analogous to the effect in a volume of gas whereby pressure changes under adiabatic conditions produce heating and cooling. In the case of solids, adiabatic thermal effects due to the application of mechanical loads produce temperature changes that are very small; typically 0.3°K for a metal subjected to a change in stress from zero to its elastic limit. Kelvin (1853) developed a basic relationship between the temperature change in a homogeneous, isotropic solid subjected to a mechanical load and the change in the sum of the principal stresses expressed as:

\[ \Delta T \propto T \Delta S \]  

(1.3.2)

Tensile changes in stress produce cooling and compressive changes, heating effects. Rocca and Bever (1950) observed the thermoelastic effect experimentally in iron and nickel using thermocouples. Subsequently a non-contacting single point infra-red radiometer was used by Belgen (1968) to investigate the influence of non-adiabatic effects, spatial resolution, optical collection efficiency, and emissivity on the thermoelastic effect observed in metals. Mountain and Webber (1978) developed a two dimensional scanning radiometer which was capable of measuring the stress distribution on a test object subjected to cyclic loading.

The relationship of the temperature change, \( T \), caused by a change in the stress in a homogeneous, isotropic, elastic solid has been derived by Stanley and Chan (1985):

\[ \Delta T = - K_m T \Delta S \]  

(1.3.3)

where

- \( T \) is the absolute temperature
- \( S \) is the sum of the principal stresses \( (\sigma_1 + \sigma_2) \)

The thermoelastic constant of a material \( K_m \) can be expressed as

\[ K_m = \frac{\alpha}{pC} \]  

(1.3.4)

where

- \( \alpha \) is the linear coefficient of expansion
Figure 2. SPATE instrumentation for thermoelastic stress analysis showing scanner head (A) and equipment rack (B)
p is the density
C is the specific heat

Significant non-linearities are introduced into the relationship between the thermoelastic response and applied load when materials are loaded plastically (Stanley & Chan 1985).

Although the thermoelastic constant \( K_m \) is considered to be independent of the mean stress level, Webber (1985) has suggested that a variation of \( K_m \) may occur in some anisotropic materials. If adiabatic conditions are to be achieved experimentally so that equation (1.3.3) is valid loads must be applied to the test specimen dynamically and at a frequency that is sufficiently high to prevent non-adiabatic effects. Such effects may be caused by a decrease in the thermoelastic temperature change due to diffusion through the substrate or environmental radiation. This effectively limits the minimum specimen loading frequency to approximately 10Hz for non-metals.

It is evident from equation (1.3.2) that the sum of the principal stresses \( (\sigma_1 + \sigma_2) \) on the surface of a specimen can be calculated from thermoelastic measurements, however, it is unfortunate that the magnitude and direction of the principal stresses \( (\sigma_1, \sigma_2 \text{ and } \Theta) \) cannot be resolved.

A full-field stress analysis instrument utilising the thermoelastic effect has been developed commercially (SPATE 9000; Ometron Ltd., London, England (2)). The instrument consists of a scanning head containing an infra-red detector which tracks over a specimen surface in a raster-like manner (Figure 2.) taking temperature readings at predetermined intervals. Maximum resolution is achieved by taking readings 150 \( \mu \text{m} \) apart. The phase of the signal from the IR detector is correlated electronically with a reference waveform obtained from a load cell or strain gauge attached to the specimen. This enables the instrument to determine thermoelastic temperature changes attributable to the change in stress whilst rejecting ambient temperature variations. Temperature changes of 0.001° C can be detected giving a sensitivity of 1Nmm\(^{-2}\) on aluminium for example.

The SPATE system has been used in biomechanics to determine variations in the stress distribution of different types of artificial hip prosthesis mounted in bone (Duncan &
Mackenzie 1989 and Duncan & Nicol 1985). Duncan and Mackenzie reported that advantages of thermoelastic stress analysis in comparison to conventional experimental and finite element methods included contact-free, high resolution imaging of structures that had a complex geometry. These authors derived a value for the thermoelastic constant \( K_m \) of bone by comparing the thermoelastic behaviour of a disc of bone subjected to dynamic loading with stresses calculated using classical elastic theory. They concluded that fresh bone gave a stress related thermal emission that was detectable by SPATE and had a stress distribution comparable to that obtainable by conventional methods. To-date, thermoelastic stress analysis has not been applied to analyse the distribution of stresses in human teeth subjected to a range of occlusal loads in-vitro.

1.3.3 Discrete Field Analysis Methods

1.3.3.1 Strain Gauges and Applications

Mechanical strain expressed as the ratio of the change in length to the original length \( \ell / \ell_0 \) can be recorded using a gauge mounted on two reference points on a specimen. The required sensitivity, range and accuracy are important parameters of such a strain gauge and influence the type chosen for a specific application. As strain gauges have a finite length a measurement error will be introduced if the strain field is not uniform and changes over the gauge length.

There are four main types of strain gauge:

1) **Mechanical**: In this type of gauge a displacement is measured mechanically by a system of levers.

2) **Optical**: Optical gauges utilise diffraction and interference patterns produced by shining monochromatic light through a slit whose width varies with displacement.

3) **Acoustical**: Acoustic gauges determine the resonant frequency of a vibrating wire.

4) **Electrical**: Small displacements can be converted into changes in capacitance, inductance or resistance.

Electrical resistance gauges are the commonest type of strain gauge in use today. They
are available in a range of design configurations and aspects of their construction and application relevant to their use for measuring strains on biological tissues including teeth are discussed in Appendix I.

1.3.3.2 Use of foil resistance gauges for strain measurement in dental research.

The use of electrical resistance strain gauges to measure mechanical strains on dentures, bridgework and teeth subjected to mechanical loads in-vitro has been described by a number of workers (Morris, Nowara & Rakowski, 1980; Randow, 1986; Glantz & Stafford, 1980; Craig & Peyton, 1967, Morin et al., 1988[a]; Donly et al., 1989). There appears to have been little published work describing their use in-vivo (Stafford & Glantz, 1991). As only single element gauges have been used in most of these studies it was not possible to calculate the magnitude and direction of the principal strains ($\epsilon_1$, $\epsilon_2$ and $\Theta$). Glanz and Stafford (1991) discussed the difficulties in achieving satisfactory environmental protection when using strain gauges mounted intra-orally in patients' dentures. In this investigation strain measurements were made at intervals over a period of twelve months. Although potential applications for rosette strain gauges were discussed by the authors only single element gauges were used in the study.

A brittle coating was used by Craig and Peyton (1965) to identify areas of high tensile strain on bridgework that had been subjected to a range of static loads. In a subsequent study (Craig & Peyton, 1967) strain gauges were applied to areas of high stress identified by the brittle coating to provide more detailed data on the magnitude of strains. In this study it was reported that strains decreased proportionally with the distance between the gauge and applied load. Use of rosette gauges would have enabled the magnitude and directions of the principal strains to be calculated.

Leary, Jensen and Sheth (1989) used strain gauges in an in-vitro study to measure the transference of loads through the roots of teeth which had been restored with posts. It was evident in this investigation that the point of load application and the specimen orientation would have produced a complex, multi-axial stress field and it would not have been possible to accurately resolve the strains with the single element strain gauge used.
Inaccuracies in the strains measured using gauges in this and similar studies could be introduced by variations in gauge position, gauge orientation and the point of load application between different specimens. This would result in significant differences in measured strains which were not attributable to changes in the main experimental variables.

Cuspal strain was measured in an in-vitro study by Morin, DeLong and Douglas (1984). Strains were measured using single element gauges bonded to the buccal cusps of extracted premolars prepared with mesio-occluso-distal cavities (MOD) and subjected to a load of 224 N. The procedure was repeated following restoration of the specimens with amalgam and composite materials. In this investigation a dental restorative resin was used to bond the gauges to the enamel surface. This is inappropriate as the adhesives used to bond strain gauges should be carefully chosen to match the mechanical properties of the carrier material.

Three parameters were calculated from the strain gauge measurements:

i) RS - Relative stiffness =

\[
\frac{\text{maximum strain in a sound tooth}}{\text{maximum strain in test tooth}}
\]  

(1.3.5)

ii) RD - Relative deformation =

\[
\frac{1}{RS}
\]  

(1.3.6)

iii) SR - Stiffness ratio =

\[
\frac{RS \text{ restored} - RS \text{ MOD cavity}}{RS \text{ sound tooth} - RS \text{ MOD cavity}} \times 100
\]  

(1.3.7)

The aim was to determine if the stiffness of the tooth under load varied with the presence and type of restoration. Although gauge position and orientation remained constant throughout the restorative procedures variation in loading position would have resulted in a difference in strain not attributable to the presence of a restoration. Use of rosette strain gauges would have enabled the direction of the principal strains to be determined and variations in load position corrected.

Further work carried out by these authors (Morin et al., 1988 [a],[b]) attempted to
measure strains on the buccal and lingual cusps of extracted premolars prepared with increasing sizes of mesio-occluso-distal cavities and restored with acid etched composite restorations. Their results showed high standard deviations for the mean strains although trends suggested that the strains were greater in prepared but unrestored teeth.

Causton, Miller and Sefton (1985) measured strain indirectly using a dial gauge to determine cuspal movement during polymerisation of bonded light curing composite materials placed in MOD cavities in extracted teeth. The step-like movement described may have been attributable to the tip of the dial gauge slipping down the cusp rather than actual cusp movement. Jensen and Chan (1985) reported a decrease in bucco-lingual cusp width of 33-45µm in premolar teeth during polymerisation of bonded composite restorations. In a second experiment single element strain gauges were bonded to the buccal cusps of extracted premolar teeth using a light curing resin. It was observed that significant tensile strains were produced during polymerisation of a light curing composite restoration placed as a single increment. It is important to take into consideration that in addition to the mechanical strains resulting from polymerisation stresses inaccuracies may be introduced by apparent changes in strain. These may be caused by continued polymerisation of the resin used to bond the gauges to the tooth or a local rise in temperature attributable to heat from the light source or exothermic changes during polymerisation of the composite. Temperature related apparent changes in strain can be largely eliminated by connecting the strain gauges in a Wheatstone bridge circuit with a temperature compensating dummy gauge.

Changes in cuspal strain and displacement were monitored simultaneously during polymerisation of bonded composite restorations in extracted teeth by Meredith and Setchell (1987). A direct correlation was observed between cuspal strain and displacement. It was also reported that the strains produced during polymerisation could be reduced by up to 25% by preparation and restoration of a mesio-distal slot in the restoration.

Donly et al. (1989) observed that a reduction in the volume of composite in a restoration achieved by the use of glass inserts reduced the level of cuspal strain during polymerisation.
1.3.3.3 Summary of experimental stress analysis methods and their application in dental research

Experimental stress analysis techniques can be used to measure the strains produced on the surface of a structure or model during application of a load. The magnitude of the principal strains and their direction with respect to an arbitrary axis need to be determined at every point on a specimen surface in order to fully resolve the stress distribution. In practical terms, those regions exhibiting the highest strains are of most interest and can be identified using a full-field technique. Full-field analysis methods can be used to determine the strain over part or all of a specimen surface but suffer from a number of practical disadvantages and may not provide sufficient data to enable complete resolution of the principal strains; brittle coatings fail by cracking in tension at a threshold strain making it difficult to determine the magnitude of strains in areas that have not cracked. A photoelastic model of a structure may be difficult to fabricate and complete resolution of the stress field may be complex. Strain gauges measure the strain between two points on a specimen surface and the magnitude and direction of the principal strains can be calculated by using three gauges mounted at fixed inter-element angles. It is therefore common practice in engineering to use a full-field technique to identify regions of high strain and then to apply gauges to enable resolution of the principal strains.

Craig and Peyton (1965, 1967) appear to be the only workers in dental research who have combined the use of a full-field technique (brittle coating) to identify regions of high strain on bridgework and then applied strain gauges to obtain more accurate information. Strain gauges have been used to measure strains on intact and restored teeth, posts and bridgework subjected to a range of mechanical loads. Strains produced in the surrounding tooth structure during polymerisation of enamel etched and dentine bonded composite restorations have been measured crudely as cuspal deflections and more accurately with strain gauges. Unfortunately, all published in-vitro and in-vivo methods have used single element gauges which has not enabled principal strains to be resolved.

Areas of high stress and potential failure in a structure subjected to a load can be identified by a range of experimental stress analysis techniques. Strain gauges although probably the most sensitive and accurate method only measure strains over a discrete
On large specimens and those having a rapidly changing stress field it is practicable to gauge areas of high stress that have been identified by using a coarser full-field technique i.e. a brittle lacquer. Although teeth are relatively small, their complex geometry and relatively high loads are likely to create high stress gradients on their surface and use of a full-field technique prior to the application of gauges should be considered.

The thermoelastic stress analysis technique has been described and appears to offer the sensitivity, resolution and accuracy necessary for measurement of surface stresses on teeth. However, potential disadvantages of this method include; the need for relatively high frequency dynamic loading, and measurements made as the sum of the principal stresses. It is proposed that this method is applied as a full-field technique to study the pattern of stress distribution on the surface of natural and restored teeth subjected to dynamic loading and identify areas of high stress suitable for gauge placement.

Applications using rosette type gauges to measure principal strains have been well documented in fields of mechanical engineering and biomechanics. It is uncertain, especially in view of the potential errors and inaccuracies associated with the use of single element gauges why rosettes have been used so infrequently in dental research. Therefore, it is proposed that rosette type gauges are used to resolve the principal strains in an in-vitro analysis of the strains in natural and restored teeth.

1.4 Numerical Stress Analysis Techniques

1.4.1 Introduction

Experimental stress analysis methods provide data regarding the magnitude and distribution of stresses on the surface of a structure. However, it is often necessary to obtain information about stresses which cannot be measured experimentally, at material interfaces or within a specimen for example. A knowledge of the stress distribution in a structure can also be useful in the design stage, prior to construction to enable regions of high stress and potential failure to be identified. The stresses in a structure of simple geometry which is subjected to a straightforward load can often be calculated directly. However, complex geometry or material properties can make this extremely difficult and a mathematical model of the structure may have to be formulated using a more
sophisticated numerical method such as finite differences or finite elements.

The evolution of the finite element method and its application to engineering problems has been described by Huebner (1974). Finite element analysis is a numerical method used to calculate the behaviour of a real structure by solving a set of equations which describe an idealised model. Such a model is constructed in two or three dimensions by dividing the structure into a series of discrete or 'finite' elements joined at nodes to form a mesh. Each element is defined by its boundary geometry, material properties and other basic parameters. These elements have a basic geometry which may be simple (e.g. triangular or quadratic) or more complex (e.g. parabolic - often termed higher order elements; referring to the level of integration used in their solution).

The accuracy of a finite element analysis can be difficult to establish but may be considered in simple terms to be the inverse of the differences between the results of the numerical and experimental methods for the same structure. Accuracy may be enhanced by increasing the number of simple elements used to represent the model or by using a smaller number of more complex higher order elements which more closely represent the actual structural geometry.

Selection of an appropriate element type and grading a mesh to obtain the optimum element density in the regions of interest is an art which often reflects the experience of the user. Meshing may be performed manually (i.e. entering the co-ordinates for each node and providing the connectivity between nodes to form elements) or more rapidly by an automatic technique which can produce regular or irregular meshes. Automatic mesh generators such as DeLauney and the advancing front method (LUSAS v.10.3, Theory Manual, 1991) produce meshes which sometimes appear irregular because in addition to the mesh density which has been specified by the user to highlight regions of interest the density in other areas reflects the way the mesh generator has been able to fill in the gaps. In general an irregular mesh density may influence the solution time but should not affect the accuracy providing the element shape is not distorted excessively (i.e. triangles do not appear as knife-point elements).

A finite element analysis may be performed in two or three dimensions. Two dimensional elastic problems may be plane strain or plane stress. In both problems the
displacement field is given by the displacements $u$ and $v$ in the directions of the $x$ and $y$ axes. The only strains and stresses considered are the three components in the $x$-$y$ plane. In the case of plane stress all other components of stress are zero and therefore do not contribute to the internal work. In plane strain the stress in a plane perpendicular to the $x$-$y$ plane is not zero. However, the strain in that direction is zero and therefore the stress does not contribute to the internal work. A structure such as a thin plate may be modelled under plane stress conditions in 2-D. Plane strain conditions may be used to represent a 2-D section of a 3-D structure which has a uniform $x$, $y$ geometry and loading conditions in the $z$ plane (i.e. an extrusion).

Three dimensional modelling using shell or brick type elements can be used to represent specimens having a more complex geometry but this too can be simplified if the model is symmetrical by modelling a section of the structure and sweeping it around its axis of rotation.

Although a finite element analysis may entail the solution of many hundreds of equations the basic principle is relatively simple and this has made finite element methods appropriate to solution by computer.

In a conventional linear elastic FE analysis the material properties (Young's modulus and Poisson's ratio) are assembled into a matrix which becomes known as the global stiffness matrix ($K$). This can be related to the displacement vector ($a$) for every node in each of its degrees of freedom and a force vector ($R$);

$$ (K)(a) = (R) $$  \hspace{1cm} (1.3.8)

This can be solved to give the stresses for each element and the nodal displacements in the $x$, $y$ and if appropriate $z$ planes. A finite element analysis can therefore be conveniently divided into three stages:

a) **Pre-processing** - The geometry of a model is constructed in outline and then meshed. Various attributes, material properties, boundary conditions and load cases are assigned to the mesh and the model is checked.

b) **Numerical analysis or solution** - The stiffness matrix is assembled and
calculations of the respective nodal displacements and element stresses are performed.

c) **Post-processing** - The resulting stresses and displacements are assembled into a convenient form for analysis and interpretation; commonly as a table or contour plot of stress distributions.

Although a linear elastic analysis is often appropriate for a large number of FE problems, in practice the mechanical behaviour of a structure may not be linear and the loads applied to it (loadcases) may be dynamic rather than static. Such behaviour complicates an analysis but with the current state of the art in finite element software aspects of non-linear dynamic behaviour can be investigated; non-linearity may be material (e.g. if the material cracks or undergoes elasto-plastic deformation) or geometric (e.g. if there are large deformations or the material snaps-through). Most software packages can model linear elastic, elasto-plastic, and visco-plastic properties.

The finite element method may be used to investigate the dynamic behaviour of a structure and eigenvalue analyses are commonly performed to determine the free modes of vibration (resonance) of a structure. More complex dynamic analyses which also utilise sinewave loading and are used to investigate the free vibration of a structure are spectral and harmonic response analysis. If the effect of a more complex load such as an impact needs to be studied then a transient dynamic analysis may be performed which is effectively a large number of static analyses performed at different time steps under different loading conditions.

It is important to appreciate that finite element analysis is considered to be an approximate method and confidence in the results obtained can only be achieved by comparison with solutions obtained from other experimental or theoretical methods.

In engineering, structures often have a regular geometry and are fabricated from isotropic materials whose properties are well documented. Loading parameters are also clearly defined and the stress distribution in such structures is well suited to analysis by the finite element method. In spite of this, a number of cases have been reported in civil engineering where errors in the application of finite element analysis at the design stage has led to serious structural failure.
Analysis of stresses in biological tissues such as bone and teeth using the finite element method is attractive because of the difficulties in making experimental measurements. However, it is important to note that the irregular geometry, poorly defined material properties and complex load cases makes analysis of these structures amongst the most complex applications for the finite element method making validation by other (experimental) techniques even more essential.

With some insight Cooke (1989) stated that

'...users of FEA programs should remember that a structure is not obliged to behave the way a computer says it should, irrespective of the cost of the program, the number of digits printed in the results, or the elegance of the graphical display...'.

1.4.2 Application of the finite element method in dental research

The finite element method has found a wide variety of dental applications which may be discussed in terms of their design.

1.4.2.1 Model geometry

Two dimensional analyses have been widely used in dentistry because model generation is simple and their solution is relatively fast and efficient. However, it is not correct to assume that a two dimensional bucco-lingual slice of a tooth is representative of the structure as a whole. Sakaguchi et al. (1991) investigated cuspal deformation of premolar teeth subjected to a range of loads using a two-dimensional plane strain mesh. Use of such a mesh was justified by stating that

'...plane strain conditions are realistic for bodies of constant cross-sectional area in the z axis and subjected to loads that act only in the x and y directions...'.

It may not be appropriate to apply this assumption to teeth which do not have a uniform cross-sectional area and may be subjected to a range of loads at points along the z axis. The authors attempted to validate the numerical analysis by using gauges to measure the strains on extracted teeth subjected to loads. The magnitude and direction of the principal strains could not be resolved with their method as only single element gauges were used. This meant that there was no way of relating the measured strains to the loading position which may well have varied in the z direction. Variations in the geometry
between the finite element model and experimental specimens were another potential source of error. In spite of these problems there was a surprisingly good correlation between experimental and numerical results. One of the main conclusions of this study was that buccal and lingual cusps moved independently when loaded. This finding may have been more attributable to the design of the finite element model than the behaviour of the tooth structure. The intercuspal area of enamel in the mesh was relatively small and had a modulus only three times greater than that of the dentine. This suggests that stresses would not be distributed effectively between cusps. Interactions between analysis the analysis design and resultant stresses should have been explored by altering the material properties of the enamel and dentine. A decrease in cuspal independence resulting from an increase in the value of Young’ modulus for enamel would have indicated a strong dependence on the model design.

A number of workers have attempted to simplify three-dimensional analyses of teeth by using axisymmetric models (Farah, Hood & Craig 1974, 1975, Selna, Shillingburg & Kerr 1975). An axisymmetric mesh can be fabricated by sweeping a two dimensional outline around an axis specified by two coordinates to form a cylindrical replica of the original shape. Unfortunately teeth are not symmetrical and an axisymmetric model cannot be considered to be a true likeness. In function teeth are most commonly subjected to loads acting over small areas. Another potential disadvantage of axisymmetric models is that the load as well as the structure is often swept cylindrically. Recently, however, development of new element types (Fourier elements; LUSAS Theory Manual v.10.3, 1991) enables asymmetric loading of an axisymmetric structure.

Selna, Shillingburg and Kerr (1975) developed an axisymmetric model of an unprepared premolar tooth and described the stress distribution resulting from a load applied at two points. It was reported that refinement of the mesh to include more elements did not significantly increase accuracy.

Three dimensional finite element analyses in dental research have generally been crude using relatively few block elements to produce a mesh which is a poor representation of the geometry of a tooth. There have been three main reasons for this; firstly, it is difficult to generate the x,y,z coordinate data which accurately reflects the external geometry of a
natural tooth and represents the appropriate material boundaries. Secondly, production of a suitable, irregular mesh from co-ordinate data may be beyond the capability of automatic mesh generators currently available. Thirdly, the resulting mesh may contain several thousand elements and even if it is assumed that the analysis is static and linearly elastic it is likely to be expensive on computer time (there is an approximate third power relationship between the number of nodes and CPU time [Rooney and Steadman, 1987]). Analysis and interpretation of the results is also likely to be difficult.

Rubin et al. (1983) carried out a crude three-dimensional analysis of a first molar tooth which appeared very angular consisting of only 336 elements. Points of load application were not clearly identified and their results were inconclusive.

A comparison of two and three-dimensional meshes of a first molar was carried out by Goel, Khera and Singh (1990). It was reported that there were significantly different levels of stress between a two dimensional model composed of 28 cubic elements and two three dimensional models, one composed of 264, 8 noded, linear elements and the other 54, 20 noded, quadratic elements. This result is not altogether surprising as the two-dimensional mesh had been constructed using a different geometry from the three-dimensional meshes which were very similar, having nodes in the same position. The load case used in this investigation was rather curious; a load of 250N was applied uniformly over the occlusal surface. This cannot be considered to be representative of loads applied clinically, even during mastication.

In further studies Khera et al. (1988, 1991) investigated the effects of cavity preparation, cavity size, depth, isthmus width and the thickness of remaining dentine using three dimensional analyses. A similar load case to the previous study was applied and the authors deduced that cavity depth may be the most significant factor in cusp fracture.

A number of techniques exist for defining the external geometry and material boundaries of a tooth in a form suitable for analysis by a finite element software package. One of the most straightforward is to use photographs or sections of a model to digitise the coordinates in the x, y and z planes which are considered to most accurately represent the specimen geometry. Facilities may exist in the finite element software to enable these
coordinates to be connected to form surfaces and/or volumes. A good analysis package will include smoothing and interpolation routines to minimise angularities and sharp corners in a mesh and provide continuity at interfaces between materials.

In applications of the finite element method in dental research, meshes have often been produced by entering node co-ordinates manually (Williams & Edmundson 1984[b]), by entering a small number of coordinates (i.e. cavity and cusp width) based on measurements taken from an experimental specimen (Hickman & Jacobsen, 1990) or average dimensions taken from the literature (Langsjoen & Noble, 1982). This has resulted in meshes which are often poor representations of the true specimen geometry and may be a source of inaccuracy.

1.4.2.2 Material properties

In finite element analysis it is important to define relevant mechanical properties which can be assigned to regions of a mesh representing different materials. In a linear elastic analysis the most important properties are Young's Modulus and Poisson's ratio. All finite element software packages can analyse isotropic behaviour and some are able to model anisotropic or orthotropic materials i.e. carbon-fibre reinforced composites and concrete.

The type of analysis performed (i.e. linear or non-linear) is dependent on the response of a material to an applied load (i.e. the shape of the stress/strain curve). Limited non-linear analyses can be undertaken using the finite element method but the material properties and type of non-linearity need to be clearly defined.

Although the mechanical properties of human enamel and dentine have been carefully investigated (see later) reported values for their elastic coefficients vary widely. This may be attributed to inherent biological variations in teeth, their complex structure, variations in test methods, specimen preparation techniques and storage conditions.

There is a wide variation in the values of elastic coefficients for enamel, dentine and restorative materials applied to numerical stress analyses in dental research. A number of authors have utilised values for Young's modulus taken from published numerical analyses and not directly or indirectly from references referring to measure-
ments of this parameter which must be considered to be unsatisfactory.

The complex structure of enamel and dentine suggest that these materials do not exhibit isotropic behaviour. However, relatively little experimental data exists to define their properties clearly in terms of specimen orientation. Rasmussen (1976) has reported that dentine may behave isotropically even though it is permeated by a large number of tubules. Van Noort, Howard and Cardew (1991) performed a numerical analysis in which enamel was modelled with curvilinear anisotropy although there is little experimental evidence to support this. On the basis of micro-hardness measurements, Davidson, Hoekstra and Arends (1974) reported that enamel exhibited isotropic behaviour. As the isotropic properties of dental hard tissues are not clearly understood and finite element analysis software is only able to model simple anisotropic materials it would seem sensible to limit analyses to those of a linear elastic, isotropic nature.

Yettram, Wright and Pickard (1976) performed a two-dimensional analysis on a premolar restored with a full gold crown. The authors assumed that dentine was isotropic but considered that enamel may be orthotropic. An analysis was carried out in which isotropic elastic coefficients were applied \( (E = 46.89 \, \text{GPa}, \, v = 0.3) \) and a further analysis employing orthotropic behaviour \( (E_x = 70.3 \, \text{GPa}, \, E_y = 23.44 \, \text{GPa}, \, v_{xy} = 0.3, \, v_{yx} = 0.1) \). The orthotropic values for enamel were derived from the modulus of rigidity which was given as:

\[
G = \frac{E}{2(1+v)} \tag{1.4.1}
\]

A comparison of the results of the two analyses did not reveal significant differences in the magnitude and direction of the principal stresses. In a further study Wright and Yettram (1978) investigated how variations in values of modulus for enamel and dentine may influence the resultant stresses in an analysis of an axisymmetric model of a second premolar restored with an amalgam restoration. Their results suggested that ratio of the modulus of the materials rather than their specific values may significantly alter the resulting stresses and displacements.

If the mechanical properties of a structure have not been established, perturbation of a range of possible values to determine their influence on the results is a logical step which
unfortunately has rarely been applied to stress analyses of teeth.

1.4.2.3 Boundary conditions and interfaces

A force applied to a structure has to be opposed by an equal and opposite force if equilibrium is to be maintained. A reactionary force in a finite element analysis is provided by restraining selected nodes which form part of the mesh. Such nodes may be fixed in one or more axes, given a degree of elasticity or permitted to undergo a prescribed displacement.

In-vivo a tooth is supported by a periodontal ligament which distributes stresses to the surrounding alveolar bone. The mechanics of this highly complex support system are not clearly understood (see later) but are visco-elastic in nature. Physiological displacement of a healthy posterior tooth in its socket when subjected to normal functional loads is approximately 50-100 μm (Parfitt, 1960). The mechanism of damping and recovery is complex and at present the limitations of the finite element method are such that it is not possible to accurately model the viscoelastic behaviour of the periodontal ligament.

Nokubi et al. (1977) attempted to model the load-displacement characteristics of the periodontal membrane using a quasi-non-linear technique based on the results of an in-vivo study of tooth mobility. This was carried out using a two-dimensional model of a premolar tooth in which the root surface was surrounded by a periodontal ligament and alveolar bone. Young's modulus and Poisson's ratio for the ligament changed from 10Nmm⁻² to 1000Nmm⁻² and from 0.3 to 0.49 respectively if the equivalent stress of any elements in the ligament reached 0.001Nmm⁻². In a second part of this study Tsutsumi et al. (1977) reported that the values of elastic coefficients used for the periodontal ligament resulted in a tooth displacement one tenth of that measured in the in-vivo study. The two stage non-linear modelling of the periodontal ligament in this study was clearly an oversimplification but little consideration appears to have been given by other workers to the boundary conditions applied in finite element analyses of teeth. In such analyses a tooth is often modelled in isolation with restrained nodes forming the horizontal boundary of the root. This will result in a very 'hard' reaction to an applied load totally unrepresentative of that seen clinically. An approach which gives a more realistic answer without the complexities of the periodontal ligament is to embed the root surface in a material having
properties similar to that of bone. Restraining the nodes forming the outer boundary of the 'bone' will damp the reaction of the load applied to the tooth.

In an attempt to be able to predict final tooth position following orthodontic treatment Williams and Edmundson (1984, [a]) carried out an analysis to determine the position of the centre of instantaneous rotation of a maxillary central incisor. In the two-dimensional model, consisting of an upper incisor surrounded by a periodontal ligament and alveolar bone, values for Young's modulus and Poisson's ratio of the periodontal ligament were varied from 0.5-100Nmm$^{-2}$ and 0-0.45 respectively. The authors reported that a change in modulus resulted in a change in the centre of rotation but variation of Poisson's ratio had little effect until a value of approximately 0.4 when the analysis became numerically unstable. This occurred because as Poisson's ratio approaches 0.5 the elements comprising the material become incompressible and the analysis will fail.

Accurate modelling of interfaces between different materials has posed a significant problem in the use of the finite element method in dental research that has often been overlooked. Amalgam restorations form an interface with the remaining tooth structure across which compressive and some shear stresses but no tensile stresses will be transmitted. However, in the case of an interface between etched enamel and composite where there may be adhesion tensile stresses will also be transmitted.

In a finite element analysis interfaces between different materials usually assume perfect adhesion with elements comprising different materials being joined at common nodes. However, such a model is clearly not representative of the amalgam/tooth interface. A number of workers (Williams, Edmundson & Rees 1987, Wright & Yettram 1978) have ignored this problem although Morin et al. (1988[b]) attempted to overcome it by introducing an intermediate layer of elements of very low modulus between the restoration and tooth. Such a model cannot be considered accurate as neither compressive or tensile stresses would be realistically transmitted from tooth to restoration.

Peters and Poort (1983) modelled an amalgam/tooth interface in three ways;

i) with no connection between the restoration and tooth,

ii) with perfect bonding,
iii) with transmission of stresses in compression only.

Although specific values for the elastic modulus of the materials used were not given the authors observed that a 0.5% change in stress was the result of a 10% change in Poisson's ratio. Their results also suggested that the distribution of the principal stresses was highly dependent on the model used for the amalgam/tooth interface.

The magnitude of the stresses at the tooth/restoration interface were examined by Van Noort, Cardew and Howard (1988) in a series of two-dimensional models of a molar tooth restored with amalgam and bonded and unbonded composite restorations having a range of values for modulus. It was observed that when a tooth was subjected to a compressive load the interfacial stresses increased with the modulus of the restorative material. The authors suggested that bond failure, due to polymerisation shrinkage, may be more likely to occur when highly filled, high modulus composite materials are used to restore teeth.

Van Noort et al. (1989) also utilised the finite element method to investigate the influence of specimen geometry, loading configuration and material properties on in-vitro bond testing of composites to dentine. Discontinuities apparent in the mesh along the central y axis in the published data were reflected in the resultant stresses and indicated potential shortcoming of the model. The behaviour of the model used in this investigation was already well understood from a simple analysis of adhesive joints (Kinloch, 1980).

1.4.2.4 Load Cases

A static, linear elastic finite element analysis indicates the distribution of stresses and displacements through a structure which has been subjected to a load at a specific point in time. Applied loads are usually resolved into their x, y and z components and, depending on the flexibility of the software, may be applied at a single node, groups of nodes or all the nodes comprising a line, surface or patch on the mesh.

In the use of the finite element method for the study of stresses in teeth applied loads have almost always been static and performed as linear, elastic analyses. There has also been a wide variation in the magnitude of applied loads. Loading position on the tooth surface has shown considerable variation between different studies. Sakaguchi et al. (1991)
applied loads at a number of nodes on the occlusal surface of a mesh to study cuspal independence, however, the mesh design and low enamel:dentine elastic modulus ratio used may have influenced the results.

It is appropriate to apply concentrated or point loads in finite element analyses of teeth as similar contacts may be seen between opposing teeth in-vivo. Axi-symmetric or surface loads such as those used by Selna, Shillingburg and Kerr (1975) and Khera et al. (1988, 1991) have to be questioned, however.

Transient dynamic analysis is potentially the most useful dynamic method for investigating the distribution of stresses in teeth but it is complex and insufficient is known at present about how loads are applied between teeth in function.

In addition to structural analysis the finite element method is also widely used for the investigation of field problems (i.e. thermal, magnetic and electrical as well as mechanical loads). A relevant application of a thermal analysis in dentistry has been the simulation of polymerisation shrinkage of a composite restoration (Hickman et al., 1991). This was performed by selecting a value for the coefficient of thermal expansion of the composite that produced a 1% decrease in the volume of the restoration when the temperature was decreased by 1°C. The resultant stresses were transmitted to the remaining tooth structure. High levels of stress were observed at the enamel/restoration/dentine interface (32.35 MPa) and the authors concluded that there was a strong possibility of bond failure as the result of polymerisation stresses in a 'bonded' composite restoration. The geometry of the mesh in this investigation appeared to be extremely angular, the boundary conditions were not clearly explained and modelling a crack by subtracting elements when joint elements may have given a more realistic result must be considered to be relatively crude.

1.4.2.5 Validation

Validation should be considered to be an essential part of any finite element analysis to increase the confidence in the results obtained using the numerical method. This is performed by comparison, where possible, with results of the same problem obtained from another source (commonly experimental or theoretical).

The accuracy of commercially available finite element software packages is assessed
by performing analyses that have explicit mathematical solutions. The suitability of such benchmarks is regulated by the National Agency for Finite Element Methods (1984).

An attempt should always be made to validate numerical results when the finite element method has been applied to study the distribution of stresses in teeth as there are a large number of factors related to geometry, material properties and load cases which may contribute to potential inaccuracies. Unfortunately it appears that finite element analysis has been used in dental research largely in an attempt to eliminate the need for experimental investigations. Relatively few workers have attempted to validate the results of their numerical analyses with experimental data.

Work by DeVree, Peters and Plasschaert (1983) illustrated the importance of selection of appropriate boundary conditions and interfaces by attempting to compare results of a photoelastic analysis by Granath (1964) with a numerical analysis that had used different mesh configurations. Morin et al. (1988 [a],[b]) compared experimental results taken from strain gauge measurements on extracted teeth with results from a two-dimensional plane strain finite element model. There appeared to be reasonable correlation between experimental and theoretical results for a prepared but not a restored tooth. This may have reflected limitations in the technique used to simulate the tooth/restoration interface discussed earlier. Other errors may have been introduced because the geometry of the finite element mesh was not based on the experimental specimens and the plane of section of the model may not have been in the same plane as the experimentally measured strains.

Similar potential problems were to be found in an investigation by Sakaguchi et al. (1991) of the cuspal deflection of intact teeth. Although there were significant differences in the geometry and loading configurations between the experimental and theoretical models there was a surprisingly good correlation in the strains recorded using the two techniques.

Another area which can be considered to be part of validation is the definition of the parameters used in an analysis. A significant number of commercially available software packages are available which utilise the finite element method; each of these has a library of element types whose solution method may vary between packages, particularly for the more complex higher order elements. It is important therefore, to define clearly in
publications the software and version used, the element type or shape and number of
nodes per element, the boundary conditions, the load cases, and the material. Unfortunately the majority of publications in this field in dentistry have failed to state
some or even all of these parameters making reproduction of the work extremely difficult.

1.4.2.6 Summary of the applications of finite element analysis in the
measurement of the stress distribution in sound and restored teeth

Finite element analysis is a numerical method in which a mathematical model of a
structure is created in two or three dimensions by defining its external geometry and
dividing it into discrete elements which are joined at nodes. Appropriate material
properties, load cases and boundary conditions are assigned to the mesh enabling a range
of static or dynamic, linear-non-linear structural stress analyses to be performed. It is an
approximate method and the accuracy of resulting stresses and displacements depends on
the quality of the input data. Validation of results by comparison with experimental or
theoretical data is also considered to be an important part of an analysis.

The finite element method has two main areas of applications: to optimise the design of
a structure, prior to its construction, and to determine the stress distribution in specimens
having a complex geometry and in which purely experimental measurements would be
difficult if not impossible to carry out. Where possible the finite element method should
be seen as part of a systematic approach to stress analysis, used in combination with
experimental methods. Unfortunately the majority of applications in dental research to
date have failed to support numerical results with experimental data. Morin et al. (1988
[a], [b]) and Sakaguchi et al. (1991) are amongst the few workers who have attempted to
validate their numerical analyses with experimental strain gauges measurements in
investigations of the stress distribution in teeth restored with composite materials and
cuspal deflection of intact teeth.

Other applications of the finite element method in dental research have been in the
measurement of stresses in teeth restored with crowns (Yettram, Wright & Pickard,
1976), amalgam and composite restorations (Wright & Yettram, 1978; Van Noort, Cardew
& Howard, 1988) and bridgework (Farah, Hood and Craig, 1975).
Finite element meshes may be three dimensional, representing the whole or a component part of a structure or two dimensional representing a section. A section of a tooth, therefore has to be modelled with care in such a way that the structure can be related to the applied load and any measured strains. Adhesive and non-adhesive material boundaries are also difficult to model accurately and such problems are evident in the dental literature.

1.4.3 Failure in Structures

Engineering structures are designed so that the ultimate load they are capable of sustaining is greater than the functional load by an amount defined as a factor of safety (Fenner, 1989). Such a consideration is equally important in biomedical applications involving the design and construction of a prosthesis but often poor knowledge of the loads in service makes accurate prediction difficult. Fracture in brittle materials and yield in ductile materials are common causes of failure. Criteria have been developed to predict the levels of stress at which fracture or yielding may occur. Prediction of yield and fracture may be made difficult by the presence of stress concentrations, localised regions of high stress which may be attributable to geometric discontinuities. Stress concentration factors \((K_f)\) have been determined for specimens of various geometries (Roark, 1985).

1.4.3.1 Yield and fracture criteria

The application of yield criteria (Hearn, 1981[b]) enables the principal stresses at a point in a material to be related to the yield stress in simple tension thus enabling prediction of the levels of stress necessary to cause failure of a structure. The Tresca or maximum shear stress criterion proposes that yielding occurs when the maximum shear stress in a complex system is equal to that in simple tension at yield:

\[ \sigma_1 = \sigma_y \quad (1.4.2) \]

Von Mises criterion (maximum shear strain energy), however proposes that yielding occurs when the maximum shear strain energy in a complex system equals that in uniaxial tension:

\[ \frac{1}{2}[(\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_3 - \sigma_1)^2] = \sigma_y^2 \quad (1.4.3) \]
Yield criteria are commonly applied to metals and materials exhibiting ductile behaviour. Brittle materials fail by fracture and the Rankine criterion (maximum principal stress) predicts that failure will occur when the magnitude of the maximum principal stress is equal to the failure stress in uniaxial tension. In practice brittle materials are much weaker in tension than compression because flaws and cracks act as local stress raisers. Mohr's fracture criterion accounts for this by allowing for different ultimate strengths in tension and compression:

\[ \frac{\sigma_t}{\sigma_{yt}} + \frac{\sigma_c}{\sigma_{yc}} = 1 \]  

(1.4.4)

In order to be able to predict how teeth may fail by fracture or yielding it is necessary to review the mechanical properties of teeth and the materials used to restore them. If failure criteria are to be applied to teeth it is necessary to establish if they behave as brittle materials.

1.5 The Physical Properties of Teeth

1.5.1 Introduction

Experimentally derived strains are often used to calculate stresses which cannot be measured directly. A knowledge of one or more of the principal elastic coefficients, Young's modulus (E), shear modulus (G) and bulk modulus (K) is necessary to do this. These constants are inter-related by Poisson's ratio (v), the relationship between lateral compressive strain and axial tensile strain;

\[ E = \frac{1 + v}{2G} \]  

(1.5.1)

\[ E = \frac{1-2v}{3K} \]  

(1.5.2)

Young's modulus, a measure of a materials' stiffness, is expressed as a ratio of stress and strain (\( \sigma/E \)) and may be derived experimentally using tension or compression tests. The applied stress can be calculated by dividing the axial load (P) by the original cross-sectional area (A_o) and strain measured using a gauge or extensometer (delta L/Lo).

An understanding of the mechanical properties of teeth and restorative materials is important to enable prediction of failure based on strain measurements. Compressive
properties of whole teeth have been investigated (Neumann & DiSalvo, 1957) although the two major components, enamel and dentine have usually been studied independently. Enamel is a highly inorganic, crystalline material composed of 94% by weight hydroxyapatite (Jenkins, 1978) which forms the outer shell of the crown of the tooth, whilst dentine contains a higher level of organic material (21% by weight) and forms the bulk of the tooth and root.

Specimen fabrication for mechanical testing of restorative materials can be carried out by moulding a test piece of the appropriate dimensions however, enamel and dentine specimens are produced by sectioning, grinding and polishing and cannot be fabricated to have an optimum test geometry.

The mechanical response of a material to stress may be brittle-elastic, elastic-plastic, viscous or visco-elastic and can be determined using a range of test methods including; quasi-static methods (tension, compression, bending and hardness) and dynamic methods (ultrasonic). In addition to variables such as specimen size and crosshead speed it is also important to consider the temperature dependence of these properties.

1.5.2 Determination of the mechanical properties of teeth using static test methods.

Reviews by Braden (1976) and Waters (1980) have discussed much of the work carried out to determine the physical properties of enamel and dentine. In most studies dentine has exhibited linear stress/strain behaviour at low strain rates but Craig and Peyton (1958[b]) observed slight creep and some hysteresis on unloading dentine specimens that had been held under a static load for twenty minutes. Assuming the test equipment used was sufficiently stable this result may be suggestive of a time dependent phenomenon. Duncanson and Korostoff (1975) also observed stress relaxation in dentine specimens loaded under constant strain. Values of Young's modulus for dentine determined using a range of static test methods and reported in the literature range from 5.85 to 19.3 GNm⁻², with a mean of 12.42 GNm⁻² (see Braden, 1976).

Dentine is a highly mineralised tissue which contains large numbers of collagen fibres, and is permeated by tubules running from the pulp to the amelo-dentinal junction.
Although such a structure is strongly suggestive of anisotropic mechanical behaviour it has been reported that specimen orientation (Craig & Peyton, 1958[b]), and tubule direction (Stanford et al. 1960, Peyton et al. 1952) do not significantly alter measured values of Young's modulus. However, small orientation dependent effects may not have been observable in these studies because of small sample numbers and difficulties in specimen preparation. It is technically difficult to measure the dimensional changes in small specimens of dentine and there has only been one reported direct measurement of Poisson's ratio of 0.014 (Haines, 1968) which is very low and may have been influenced by the specimen geometry in which dentine was constrained on three sides by enamel. Renson and Braden (1975) calculated a value for the rigidity modulus of dentine as 0.62 GNm$^{-2}$ and from this deduced a value for Poisson's ratio of 0.12. As Waters (1980) has already discussed this value appears to have been miscalculated and should be 0.23.

Temperature can significantly alter the mechanical behaviour of materials viz. the glass transition temperature of polymers. El Mowafy and Watts (1989) measured the compressive properties of dentine at a range of temperatures (23, 37, 50, 80°C) and Watts (1989) deduced a negative linear relationship between Young's modulus and temperature. The value for dentine, $-0.073\text{ GNm}^{-2}/^\circ\text{C}$, is in good agreement with the temperature coefficient of modulus for cortical bone, $-0.07\text{ GNm}^{-2}/^\circ\text{C}$ reported by Bonfield and Tully (1984). Environmental test conditions are rarely quoted in investigations of the mechanical properties of dentine and it has to be assumed that most tests have been carried out at ambient temperatures and humidities.

Enamel, a highly inorganic, crystalline material, forms a layer over the surface of the dentine comprising the crown and varying in thickness from a few microns at the cervical margin to approximately 2mm in the occlusal fossa. It is hardly surprising, therefore, that workers have reported difficulties in preparation of enamel specimens suitable for testing (Craig et al., 1961; Stanford et al., 1958; Tyldesley, 1959; Hannah, 1970). In an attempt to overcome this problem Tyldesley (1959) tested composite enamel/dentine specimens in four point bending. However the mean value for Young's modulus of enamel calculated in this study (130 GNm$^{-2}$) was very high. Values for Young's modulus of enamel reported in the literature range from 24.8 GNm$^{-2}$ (Stanford et al. 1958) to 81.0 GNm$^{-2}$ (Craig et al.
1961) with a median value of 40.6 GNm\(^{-2}\). This wide range may in part be attributable to variations in test method, specimen location and preparation.

Structurally, enamel and dentine can almost always be considered in unison; it is therefore surprising that few workers have investigated the stress/strain behaviour of whole teeth. Neumann and DiSalvo (1957) reported that sound teeth subjected to compressive loading exhibited linear stress/strain behaviour but yielded slightly prior to failure. Watts (1986) observed similar characteristics in molar teeth restored with etched enamel MO and MOD composite restorations.

1.5.3 Determination of the mechanical properties of teeth using dynamic test methods

Dynamic test methods for the determination of elastic properties can offer a number of advantages over static techniques;

a) Smaller specimens can be used especially with high frequency techniques
b) An absence of moving parts can result in higher accuracy and precision
c) Stress and strain amplitudes produced by dynamic methods are very small (often lower than 0.1MPa and \(10^{-5}\) m) and well suited to measurements of the elastic moduli of brittle materials.

The elastic coefficients of a material may be calculated from measurements of the velocity of propagation of elastic stress waves through a solid. Such waves may be longitudinal, extensional or shear. Three methods have been used to measure wave velocity in enamel; pulse echo (Barber, Lees & Rollins, 1972), interferometric (Gilmore, Pollack & Katz, 1969) and reflection ratio (Lees, 1968). The elastic coefficients of a material may be related to wave velocity by the specific acoustic impedance \(Z\);

\[
Z = \sqrt{Kp} \quad \text{and} \quad c = \frac{\sqrt{k}}{p} \quad (1.5.3 \& 1.5.4)
\]

where

\[
K = \text{Bulk modulus}
\]

\[
p = \text{Density}
\]

\[
c = \text{wave velocity}
\]
Specific acoustic impedance measurements have been used to calculate the elastic moduli of human enamel in a number of studies (Lees, 1968; Barber, Lees & Rollins, 1972; Gilmore, Pollack & Katz, 1969). Variations in values between different investigations may be attributable to the test method and frequency used, and to variations between different specimens. Values reported by Lees (1968) can be considered to be representative of values of modulus made by measurement of wave velocity in enamel:

\[
\begin{align*}
\text{Young's modulus (E)} & = 76.5 \text{ GN/m}^2 \\
\text{Bulk modulus (K)} & = 65.3 \text{ GN/m}^2 \\
\text{Shear Modulus (G)} & = 29.3 \text{ GN/m}^2 \\
\text{Poisson's ratio (v)} & = 0.30
\end{align*}
\]

The dynamic properties of materials at high frequencies cannot be considered to be directly comparable with quasi-static measurements and often result in higher values for the elastic coefficients than those derived from static measurements c.f. Young’s modulus for enamel: 76.5 GNm\(^2\) (Dynamic; Lees, 1968) and 24.8GNm\(^2\) (Static; Stanford et al., 1958).

1.5.4 Hardness and fracture characteristics of enamel and dentine

The hardness properties of enamel and dentine have been studied by a number of workers. Reasons for the popularity of this type of test method probably include ease of specimen preparation, rapidity of test and collection of data. Testing has been carried out using spherical (Renson & Braden, 1971), Vickers (Ryge, Foley & Fairhurst, 1961) and Knoop indentors (Craig & Peyton, 1958 [a]; Davidson, Hoekstra and Arends, 1974). The Knoop method (Knoop, Peters & Emerson, 1939) employs a pyramidal shaped diamond stylus which is used to indent a specimen surface. A Knoop hardness value for a material is based on the relationship between the long indentation diagonal and applied load. This has proved to be the most popular method in dental research and a range of hardness values for enamel and dentine have been reported in the literature (Braden, 1976):

\[
\begin{align*}
\text{Enamel HK} & = 272 - 440 \\
\text{Dentine HK} & = 50 - 70
\end{align*}
\]
The majority of investigations have measured the hardness of enamel and dentine on bucco-lingual sections of sound teeth. Sample preparation entails sectioning, grinding and polishing which will modify the specimen surface and resultant hardness. Specimen thickness is also of importance: Waters (1965) has shown that the ratio of specimen thickness to contact radius should be large enough to minimise stresses on adjacent surfaces: greater than eight for a spherical indenter.

El Mowafy and Watts (1989) observed a negative, linear temperature coefficient of $-0.014/°C$ for the Knoop hardness of dentine in the range 0-60°C. Micro-hardness measurements have been used to investigate if the mechanical behaviour of enamel and dentine can be related to their structure. Craig and Peyton (1958 [b]) were unable to establish a relationship between Knoop micro-hardness and the position and orientation of an indentation in enamel or dentine. In addition no significant differences were observed in hardness between teeth of different types. Interestingly, the authors observed a contraction of the short diagonal in indentations of dentine which they attributed to elastic recovery. Unfortunately this was not investigated further. In a subsequent study (Craig, Peyton & Johnson, 1961) it was found that the hardness of dentine was lowest adjacent to the pulp ($HK = 30$) or amelo-dentinal junction ($HK = 15$).

Enamel is structurally complex, consisting of hydroxyapatite crystals arranged in key-hole shaped prisms. Although this is suggestive of anisotropic behaviour, Davidson, Hoekstra and Arends, (1974) concluded that this material was mechanically isotropic on the basis of micro-hardness measurements made on the enamel surface.

The mechanical response of dentine to stress was investigated by Renson and Braden (1971) using different types of indentor (spherical, cylindrical and conical). The authors observed that when using a conical indentor, dentine behaved elastically at low loads and plastically at higher loads. Braden (1976) and Renson and Braden (1971) referred to a value for Young's modulus for dentine ($12.0 \text{ GNm}^{-2}$) deduced from indentation measurements but the method of calculation was not described.

The fracture behaviour of a material can give a useful indication of brittle or ductile characteristics. Quantitative data on the fracture of enamel and dentine has been obtained using two main methods: work of fracture and fracture toughness.
Work of fracture is defined as the work required to form a new surface of unit area ($W_f$). Tattersall et al. (1966) described a method for measurement of $W_f$ using a rectangular specimen which had a V shaped central section thus limiting catastrophic crack growth and resulting in a controlled fracture. Rasmussen et al. (1976) applied this technique to a study of enamel and dentine. In contrast to other workers Rasmussen observed anisotropic behaviour in both enamel and dentine;

<table>
<thead>
<tr>
<th>Material</th>
<th>Direction</th>
<th>$W_f$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dentine</td>
<td>(perpendicular to tubule direction)</td>
<td>$270 \text{Jm}^{-2}$</td>
</tr>
<tr>
<td>Dentine</td>
<td>(parallel to tubule direction)</td>
<td>$550 \text{Jm}^{-2}$</td>
</tr>
<tr>
<td>Enamel</td>
<td>(perpendicular to prism direction)</td>
<td>$200 \text{Jm}^{-2}$</td>
</tr>
<tr>
<td>Enamel</td>
<td>(parallel to prism direction)</td>
<td>$13 \text{Jm}^{-2}$</td>
</tr>
</tbody>
</table>

These values are small in comparison to those for ductile materials ($W_f$ for copper = $50 \text{kJm}^{-2}$). Enamel and dentine may therefore be considered to exhibit brittle behaviour. The results may have been influenced by overheating due to the use of uncooled abrasive discs during specimen preparation. In addition Rasmussen et al. (1976) failed to describe the specification of the load cell and testing machine used, a factor which Tattersal and Tappin (1966) had shown could lead to a variation in values of $W_f$.

Fracture toughness is an intrinsic property that may be used to characterise a material's resistance to fracture. Specimens of suitable shape can be loaded to determine the critical stress intensity factor, $K_c$, which gives an indication of the flaws that may be tolerable in a structure under given conditions. Although the technique has been widely used to study restorative materials few workers have attempted to measure $K_{IC}$ for enamel and dentine probably because of difficulties in specimen fabrication. Hassan and Caputo (1981) measured the fracture toughness of enamel and El Mowafy and Watts (1986) obtained a value of $3.08 \text{MNm}^{-1.5}$ for dentine using a compact specimen geometry based on that described in British Standard BS 5447:1977.

1.5.5 Summary of the physical properties of teeth

A number of methods have been used to define the physical and mechanical properties of enamel, dentine and intact teeth. The small size, irregular geometry and complex
structure of human teeth made use of conventional test methods difficult. Standard specimen geometries (El Mowafy & Watts 1986; Rasmussen et al., 1976) and testing rigs (Tyldesley, 1959; Craig & Peyton, 1958[a]; Renson & Braden, 1971) have been miniaturised to enable small enamel and dentine specimens to be studied. Reported values for the elastic coefficients of these materials have varied widely, especially between different studies (see above). This may, in part, be attributable to characteristics of the test procedure (Rasmussen 1976) as well as to the anisotropic nature of the materials.

Young's modulus and Poisson's ratio are the most useful and widely applied elastic constants in the calculation of stresses and measurement of strains in a structure. Unfortunately there are no reports of direct measurements of Poisson's ratio for enamel and only one for dentine (Haines, 1968). A technique capable of measuring Young's modulus and possibly Poisson's ratio over a small area of tooth surface would be valuable in determining small regional variations in material stiffness and studying possible anisotropy. From a review of the literature it would appear that micro-hardness measurements can be successfully made over a small, discrete area of a tooth surface (Davidson, Hoekstra and Arends, 1974).

1.6 The Mechanical Properties of Aesthetic Restorative Materials

1.6.1 Introduction

Measurement of the mechanical properties of dental restorative materials is performed for two main reasons; Firstly, a knowledge of basic parameters such as Young's modulus, hardness, strength and toughness enables comparisons to be made between different types and brands of materials. This also aids the development of new materials purported to have improved properties. Secondly, in-vitro test methods capable of predicting the clinical performance of a material could provide quantitative data that is more reliable and less expensive than a clinical trial. It has been suggested (Ferracane, 1985; Lloyd, 1982; Bryant & Mahler 1986) that in-vitro measurements of abrasion resistance, hardness, fracture toughness and flexural strength of restorative materials may be used to aid prediction of in-vivo performance.

Aesthetic restorative dental materials may be divided into two major groups according to their composition: composites and polyalkenoate cements.
Composite materials consist of a resin phase, commonly BIS-GMA or a derivative loaded with an inert, inorganic filler. The material cures by free radical addition polymerisation initiated chemically or by visible light activation of a tertiary aromatic amine (camphoroquinone). Filler particles vary in size (≈0.001-50μm) shape and quantity, (≈60-90% by weight). Composites of this type in which the filler is relatively brittle but the matrix may be ductile generally require a strong bond at the resin/filler interface to achieve high toughness, and filler particles are commonly coated with a silane coupling agent in an attempt to achieve this. Additional resins and modifiers may be added to alter the materials rheology and modify shade and opacity.

Glass ionomer cements are composed of a poly(alkenoic) acid and aluminosilicate glasses. The setting reaction on mixing the two phases is complex but basically involves dissolution of the glass surface by the acid with an exchange of metal ions to form a salt hydrogel. Modifiers such as tartaric acid may be added to control working time. Full maturation of the glass-ionomer matrix may take up to 24 hours. A recent development in this type of material has been the addition of light-activated pendant methacrylate groups to the poly(alkenoic) acid. The result is a material in which the resin phase polymerises rapidly on exposure to a visible light source but the glass ionomer reaction continues over a prolonged period (Vitrebond; 3M UK, Loughborough, England (3)).

1.6.2 Mechanical testing methods suitable for restorative materials

A number of investigations have used quasi-static measurements of the stresses needed to cause failure of specimens loaded in tension, compression or flexure to determine the physical properties of restorative materials. Variations of such test methods have included; use of a four point flexural test to overcome problems associated with geometry and brittle materials or an indirect (diametral) tensile test better suited to small specimens.

Tensile strength and elastic modulus of a range of materials including a silica reinforced resin were measured by Bowen & Rodriguez (1962) who reported a mean tensile strength and Young’s modulus for the composite material of 7MNm⁻² and 2.32Gnm⁻² respectively. These authors reported difficulties in mounting specimens without fracturing and tensile measurements have not been widely used for restorative
materials probably because of their brittle behaviour and difficulties in gripping and aligning specimens.

Flexural testing of materials in three or four point bending has been considered by a number of workers (Bryant & Mahler, 1986) to represent clinical loading conditions. Chitchumnong et al. (1989) reported no significant differences in values of Young's modulus between specimens of denture acrylic loaded in three and four point bending. However, values of flexural strength for these materials were consistently higher for three point in comparison to four point bending. This finding is rather surprising as specimens of brittle materials are often weaker in three point bending due to the presence of surface flaws and cracks.

Young's modulus for a number of composites and amalgam alloys measured by Bryant and Mahler (1986) in three point bending ranged from 4.5-18.8 and 37.8-60.1 GPa respectively. The composite materials could be divided into two groups according to filler particle size: those containing fine sized fillers (1-3μm) were found to have moduli in the range 9.5-10.8 GPa and those with larger particles (>10μm), higher values in the range 15.2-18.8 GPa. It was of interest to note that these authors used a relatively high crosshead speed (1.25mm/min) in an attempt to minimise the effects of creep, although their loading arrangement was somewhat unusual in having the specimen supported by two rollers with the load applied by a small steel bearing (1.59mm dia.). This may have resulted in bi-axial flexure of the specimen.

Bi-axial flexure testing of materials was advocated by Ban & Anusavice (1990) as they considered it may reduce the variance of fracture strength values of brittle materials. In a comparison with four point bending and a diametral tensile test the authors observed good agreement between bi-axial flexure and four point bending but much lower values for the mean strengths of materials in the diametral tensile test. The diametral tensile test has been criticised by Zidan et al. (1980) because of the complex stress distribution and possible deformation of this type of specimen prior.

It has been reported that dental composites behave as brittle materials at the temperatures, loading rates and magnitudes likely to be encountered clinically (Lloyd, 1982). The failure of this type of material has been shown to be statistically distributed as
a function of its homogeneity (Oysaed & Ruyter, 1986). The Weibull distribution (Weibull, 1951) is a statistic that has been commonly used to describe this distribution (McCabe et al., 1990), although it has been suggested that this technique may have limitations in predicting the failure of components having a complex geometry which are subjected to a multi-axial state of stress.

Compressive testing of restorative materials has not been widely carried out. Oysaed & Ruyter (1986) compared the compressive and flexural properties for a range of composite materials that had been stored under dry and wet conditions. Values for the elastic modulus of materials tested in compression were consistently lower than those observed in three point bending, 4.6-15.2 GPa and 6.0-19.1 Gpa respectively. Calculated values of Young’s modulus for a material will depend on the method used to determine the slope of stress strain curves and may be a potential source of error. Compressive strength, however, was consistently higher than flexural strength (283-365 MPa and 69-156 MPa) supporting the contention that composite materials may exhibit brittle behaviour.

Dynamic methods for determination of the mechanical properties of composite materials have been used by a number of workers. These have included use of a torsion pendulum (Whiting & Jacobsen, 1980[a]), ultrasonic techniques (Whiting & Jacobsen, 1980[b], Nakayama et al., 1974) and dynamic resonance methods (Braem et al., 1986). Braem et al. (1989) used a resonance method in combination with a three point flexural test to investigate the influence of variations in filler loading on the elastic modulus of a composite material. An exponential relationship was observed between filler loading and Young’s modulus. Calculated values of modulus were approximately 70% greater for the dynamic method. The ranges of values for Young’s modulus of composites reported by Whiting & Jacobsen (1980[b]) and Nakayama et al. (1974) using dynamic methods were also high at 7.4-25.55 GPa and 33-38 GPa respectively. Nakayama attributed this to reduced porosity and strain hardening of the resins by pressure cycling during curing. It is likely that the high stress rates and low displacements used in ultrasonic techniques also contributed to these findings. Dynamic methods are also useful in the ability to calculate values for shear and bulk modulus and Poisson’s ratio in addition to Young’s modulus. Determination of a value of Poisson’s ratio for composites using static test methods is
difficult and does not appear to have been reported in the literature. Values quoted using dynamic methods range from 0.23-0.32 (Whiting & Jacobsen, 1980[b]).

Mechanical properties of composite materials are influenced by their composition, the degree of polymerisation and environmental conditions especially temperature and humidity.

1.6.3 The influence of material composition and the degree of polymer conversion on the mechanical properties of restorative composites.

Bowen (1956) and Bowen and Rodriguez (1962) have shown that an increase in the filler fraction of a dental composite results in an increase in its elastic modulus and tensile strength. Chung (1990) reported a linear relationship between filler loading (66.4 - 85.2%), Knoop micro-hardness and diametral tensile strength. Boyer, Chalkley and Chan (1982) also observed a positive correlation between filler loading and elastic modulus, proportional limit and the tensile strength. It therefore seems that an increase in filler loading will result in a stiffer composite of higher strength. Surface treatment of the filler with a silane coupling agent has also been shown to result in an increase in yield stress and stress intensity factor, $K_{IC}$, calculated for a range of composites of different filler loadings (7, 15, 26, 41% by volume) using the double torsion test (Davis & Waters, 1989). Chemically curing restorative composites have largely been superseded by visible light curing materials offering improved clinical handling. However, concern has been expressed that these materials may not be fully polymerised following placement and a number of in-vitro studies have been carried out to determine the level of polymer conversion and curing depth of these materials (Rueggberg & Craig, 1988; Ferracane, 1985; Watts, Amer & Combe, 1987; Von Fraunhofer & Curtis, 1989).

Chung (1990) reported a level of polymer conversion for visible light cured composite materials of between 43.5 and 85.2% using Fourier transform infra-red spectroscopy (FTIR) but did not establish a relationship between the degree of conversion of a composite and its elastic modulus or Knoop micro-hardness. However, Rueggberg and Craig (1988) and Ferracane (1985) established a linear relationship between Knoop micro-hardness and the level of polymer conversion using FTIR and both authors suggested that relative hardness was a good indicator of the level of polymer conversion in a composite
but Ferracane (1985) reported significant differences in values of absolute hardness between different materials. Thus a single value of hardness could not be used to predict the degree of polymerisation of a material, indeed the hardness value deduced from an indentation of a composite may depend on the resin present, its level of polymerisation, the quantity and type of filler and the presence of a coupling agent. Flow of the resin due to visco-elastic recovery may also lead to distortion of the indentation.

Von Fraunhofer & Curtis (1989) considered surface hardness to be an important parameter that may influence the wear of composite restorations and observed a range of values for different composites which they attributed to a partially set surface layer resulting from oxygen inhibition of polymerisation of the resin phase. Watts, Amer & Combe (1987) measured a rapid increase in surface hardness for a composite in the first hour following exposure to a visible light source. A gradual increase in hardness continued for up to a week and lower values were obtained for specimens stored in water.

Depth of cure is an important clinical consideration in the use of light curing materials and the degree of polymer conversion will be influenced by a number of factors: the power and spectral distribution of the light source, distance of the light source from the material, the thickness and shade of the material, the reflectance and absorption of cavity walls and the levels of photo-initiator present. Curing depth has been determined in-vitro by micro-hardness measurements of composite specimens of varying thickness. Onose et al. (1985), who measured hardness at different depths by sectioning specimens longitudinally observed the highest hardness at 0.5-1.0 mm which decreased with increasing depth. The heat generated by the sectioning and polishing procedures in this study may have influenced the results.

1.6.4 The influence of environmental conditions on the mechanical properties of composite materials

In-vitro testing of restorations should be performed under environmental conditions representative of those experienced by a restorative material in clinical service. The physical properties of polymer based materials are temperature dependent undergoing a change from glassy to rubber-like behaviour at the glass transition temperature (Tg).
Fortunately this is higher than intra-oral temperatures (=32°C, with a range of 2-50° for food and drink) for the polymers used in dental restorations (93°C for the resin phase of P50 (11)). Humidity is also important and has a significant effect on the physical properties of poly-alkenoic acid based materials during setting. The influence of temperature and humidity on the mechanical properties of aesthetic restorative materials has been studied by a number of workers.

A decrease in Young's modulus and compressive and yield strength was observed for a number of composite materials tested over a temperature range of 2-80°C (Draughn, 1981). Oysaed and Ruyter (1986) also reported a decrease in modulus and tensile strength of composites stored in water for three months at 37°C prior to testing. Similar results were obtained by Soderholm and Roberts (1990). The tensile strength of three materials stored in water but then allowed to dry out was higher than materials stored and tested wet, but lower than materials kept dry. A plausible explanation for this behaviour was that hoop stresses resulting from polymerisation shrinkage formed in the resin surrounding the filler particles and these were relieved by the addition of water which acted as a plasticiser.

1.6.5 Summary of the mechanical properties of composite materials

Nakayama et al. (1974) have suggested that the mechanical properties of a restorative material should match the properties of enamel and dentine. This is difficult to achieve using a single restorative material as the enamel:dentine modulus ratio is approximately 6:1 and it is evident from the literature that the elastic coefficients of most of the currently available composites are comparable to the range of reported values for Young's modulus of dentine but are considerably lower than those for enamel. The significance of this is uncertain but Bryant & Mahler (1986) have suggested that the low modulus of aesthetic restorative materials may result in bending and flexure of the restored tooth under occlusal stresses which could contribute to the post-operative sensitivity observable clinically with this type of restoration.

Although it has been shown that specimen storage and testing temperature and humidity affect the mechanical properties of composite materials (Draughn, 1981) the majority of studies have not reported values for these variables.
A range of test methods, specimen geometries and loading conditions have been used to measure the elastic properties and strength of aesthetic restorative materials. No single method seems to be entirely satisfactory and compressive and diametral tensile tests may not be the most suitable for materials exhibiting brittle characteristics. Flexural methods used to determine values for Young's modulus and flexural strength have produced satisfactory results (Bryant and Mahler, 1986) and use of a four point bending configuration should be considered to achieve a uniform bending moment. If problems associated with specimen mounting can be overcome, tensile methods also appear to be satisfactory (Bowen & Rodriguez, 1962).

For composites, as with many other materials, values for elastic coefficients calculated using ultrasonic techniques are significantly higher than those derived using static methods. Although there is no doubt that dynamic test methods have a place, it is suggested that they should be supported with static test data where possible. The vast range of restorative composite materials available makes comparison between different studies extremely difficult. In an attempt to identify those variables between different test centres not attributable to material selection or test method McCabe et al. (1990) used a Weibull distribution to determine the probability of failure plotted against compressive and flexural stresses for a range of materials tested in a multicentre, round robin type study. It was observed that flexural tests for composites gave very consistent results.

In view of the difficulties in trying to predict the mechanical properties of materials such as composites because of the number of variables associated with the material, specimen fabrication and storage and the test method, it is proposed that static test methods should be used to determine the mechanical properties of aesthetic restorative materials used in this investigation. This should include a combination of tensile and flexural tests performed to determine values of elastic modulus and, if possible, Poisson's ratio.

1.7 In-vivo and In-vitro loading of teeth

1.7.1 Introduction

Mechanical loads applied to teeth in-vivo may be defined by a number of parameters;
amplitude, stress rate, waveform, frequency, and position. Investigation of these characteristics has been carried out clinically in a number of ways;

i) by measurement of occlusal forces between teeth.

ii) the study of maxillo-mandibular relationships during function.

iii) investigation of the electrical activity in the muscles associated with mandibular movements (electromyography).

iv) determination of the role of the tissues supporting teeth in the distribution of applied loads.

Teeth are subjected to a range of loads in function. Impact loads having a high amplitude and rate of application may result from trauma and can cause fracture of the tooth and or the surrounding bone. Loads between opposing teeth during activities such as mastication, swallowing or speech maybe of much lower amplitude and act on a number of teeth simultaneously. In parafunctional activities such as bruxing, loading may be of relatively high magnitude but act on few teeth. The study of how teeth meet (occlusion), the loads developed, and their distribution through the teeth and surrounding tissues has been made difficult by the wide range of variables. Such knowledge could lead to an increased understanding of how and why teeth and restorations fail clinically.

A number of workers (Bell, Smith & DePont, 1982; Hiatt, 1973) have attempted to correlate clinical tooth and restoration fracture with applied loads. However, it is often difficult in retrospective studies such as these to determine the aetiological factors that have led to failure and which may be due to a combination of adverse loading, poor restoration design or an inappropriately used restorative material.

The ability in-vitro to define and control variables associated with loading and restoration of specimens should enable a more accurate assessment to be made of those factors associated with the failure of sound and restored teeth. A number of in-vitro investigations have been carried out which may be divided broadly into two groups: studies in which loads are applied to prepared and restored extracted human teeth until fracture occurs and secondly, those in which the effects of loads applied to teeth are measured by transducers. Such studies suffer from the often appropriate criticism that in-
vitro experimental models cannot be related to the clinical environment. The first stage in the development of such a model is to establish the primary experimental variables. A brief review of clinical techniques used to determine in-vivo loading conditions and their application to in-vitro studies highlights some of the difficulties.

1.7.2 In-vivo measurement of occlusal forces

In order to be able to measure interocclusal forces in-vivo a load transducer has to be placed between opposing teeth. The resultant mandibular displacement is likely to lead to the application of an uncharacteristic load being applied to the transducer and Nyquist and Owall (1968) have shown that application of such a method may alter the normal range of biting forces. A number of workers have fitted strain transducers inside crowns (Anderson, 1953; Nyquist & Owall 1968), bridgework (Scott & Ash 1966) and dentures (Howell & Brudevold 1950) so that intercuspal contacts could be achieved without an abnormal mandibular displacement. However, it cannot be assumed that the stress distribution within a crown or piece of bridgework will be the same as that in the natural dentition.

Anderson (1956[a]) determined masticatory loads in-vivo by measuring the deflection of a mild steel beam fitted inside a crown using a strain gauge. However, by using only a single gauge Anderson could not resolve the magnitude and direction of the principal strains. A gradual increase in force observed during testing in this investigation was attributed to changes in chewing pattern but could have been the result of temperature induced strain gauge drift as no attempt appears to have been made to compensate for 'apparent' changes in strain due to temperature fluctuations.

Nyquist and Owall (1968) discussed the difficulties of measuring occlusal loads using single element piezo-ceramic transducers mounted inside crowns. In their investigation the output voltage of the transducer was directly proportional to the applied load but variations in the mounting angle of the transducer between the crown and longitudinal root axis made comparison of results between different patients impossible. In a subsequent paper Ahlgren and Owall (1970) attempted simultaneous measurement of mandibular position, electrical muscle activity and chewing force. An uncalibrated load
<table>
<thead>
<tr>
<th>Author</th>
<th>Year of publication</th>
<th>Measurement method</th>
<th>Transducer placement</th>
<th>Number of subjects</th>
<th>Parameter</th>
<th>Tooth position</th>
<th>Dentate/ edentulous</th>
<th>Force</th>
</tr>
</thead>
<tbody>
<tr>
<td>Howell &amp; Manly</td>
<td>1948</td>
<td>Strain gauge</td>
<td>Inter-occlusal</td>
<td>4</td>
<td>Biting force</td>
<td>Lower molar</td>
<td>Dentate</td>
<td>413-898</td>
</tr>
<tr>
<td>Manly &amp; Winton</td>
<td>1951</td>
<td>Strain gauge</td>
<td>Inter-occlusal</td>
<td>100</td>
<td>Biting force</td>
<td>Lower molar</td>
<td>Edentulous</td>
<td>9-445</td>
</tr>
<tr>
<td>Howell &amp; Brudevold</td>
<td>1950</td>
<td>Strain gauge</td>
<td>Intra-coronal</td>
<td>NG</td>
<td>Mastication</td>
<td>Lower molar</td>
<td>Edentulous</td>
<td>4.5-377</td>
</tr>
<tr>
<td>Anderson</td>
<td>1953</td>
<td>Strain gauge</td>
<td>Intra-coronal</td>
<td>4</td>
<td>Biting force</td>
<td>Lower molar</td>
<td>Dentate</td>
<td>0.4-17.5N/m</td>
</tr>
<tr>
<td>Anderson</td>
<td>1956 a,b</td>
<td>Strain gauge</td>
<td>Intra-coronal</td>
<td>2</td>
<td>Mastication</td>
<td>Lower molar</td>
<td>Dentate</td>
<td>72-145</td>
</tr>
<tr>
<td>Atkinson &amp; Shepherd</td>
<td>1967</td>
<td>Strain Gauge</td>
<td>Intra-coronal</td>
<td>2</td>
<td>Biting force</td>
<td>Lower molar</td>
<td>Edentulous</td>
<td>70</td>
</tr>
<tr>
<td>Nyquist &amp; Owall</td>
<td>1968</td>
<td>Piezo-transducer</td>
<td>Intra-coronal</td>
<td>2</td>
<td>Biting force</td>
<td>Lower molar</td>
<td>Dentate</td>
<td>N</td>
</tr>
<tr>
<td>Ahlgren &amp; Owall</td>
<td>1970</td>
<td>Piezo-transducer</td>
<td>Intra-coronal</td>
<td>2</td>
<td>Mastication</td>
<td>Lower molar</td>
<td>Dentate</td>
<td>N</td>
</tr>
<tr>
<td>Graf, Grassl &amp;</td>
<td>1974</td>
<td>3-element Piezo</td>
<td>Intra-occlusal</td>
<td>1</td>
<td>Mastication</td>
<td>Lower molar</td>
<td>Dentate</td>
<td>N</td>
</tr>
<tr>
<td>Aerberhard</td>
<td></td>
<td>transducer</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>x - 2-13</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>y - 5-20</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>z - 2-40</td>
</tr>
<tr>
<td>Lundgren &amp; Laurell</td>
<td>1984</td>
<td>Strain gauge x 4</td>
<td>Intra-occlusal</td>
<td>1</td>
<td>Biting force</td>
<td>Upper anterior</td>
<td>Dentate</td>
<td>20-25</td>
</tr>
<tr>
<td>Neill et al.</td>
<td>1989</td>
<td>Strain gauge</td>
<td>Inter-occlusal</td>
<td>10</td>
<td>Mastication</td>
<td>Upper anterior</td>
<td>Dentate</td>
<td>46-65</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>2</td>
<td></td>
<td></td>
<td>Edentulous</td>
<td>34-45</td>
</tr>
</tbody>
</table>

Table 1. Summary of in-vivo methods used for the determination of occlusal forces
transducer similar to that in the previous study was used but no quantitative data was obtained.

Resolution of occlusal forces into three components (axial \([z]\), mesio-distal \([x]\) and bucco-lingual \([y]\)) was made possible by the development of a three element piezo-ceramic transducer built into the pontic of a bridge replacing a lower first molar (Graf, Grassl and Aeberhard, 1974). Their results highlight the inability of earlier work, which had used single element transducers, to show that masticatory loading acting on a single tooth (during mastication - of medium hard Swiss rye bread) can be resolved in three planes.

Although the majority of studies have attempted to measure occlusal loads between opposing teeth, contacts commonly occur between a number of teeth simultaneously. Lundgren and Laurell (1984) used four strain gauge transducers, one mounted in a pontic space in each quadrant of a cross arch bridge, to record loads resulting from simultaneous occlusal contacts. Loads were measured on all strain gauges simultaneously and the results indicated the presence of multiple contacts which varied in magnitude. This may have been due to the transducer mounting and design which limited load measurements to those in an axial direction. It was difficult to establish a relationship between the loads recorded by the individual transducers in this study.

Recent developments in flexible printed circuit (PCB) and piezo-film technology have led to the fabrication of load transducers to measure occlusal forces from load sensitive, electrically conducting films (Maness and Podoloff, 1989). However the width of the PCB tracks (approximately 500 μm) and the overall film thickness (approximately 150 μm) has enabled only relatively crude measurements of the size and distribution of occlusal contacts to be made.

It has been shown that the magnitude of occlusal forces between teeth may vary with age (biting forces increase in children with age until adult; Worner and Anderson, 1944), tooth position (occlusal forces are higher in posterior teeth; Howell & Manly, 1948) and in dentate and edentulous patients (loads are usually lower in the edentulous; Howell & Brudevold, 1950).

Table 1 summarises the wide range of occlusal forces reported in the literature. It has
Figure 3. Representative occlusal loading curves taken from three studies
also been reported (Lindquist & Ringquist, 1973), that there is no significant difference in biting force between normal patients and those who exhibit a bruxing habit. The form of the chewing cycle, speed of movement of the mandible and chewing rate are also features of masticatory function that have been investigated and are the subject of a detailed review by Bates, Stafford and Harrison (1975, [a] & [b]). However, there appear to have been few investigations of loading rate and contact duration. Working side and non-working side occlusal contacts of 0.4 and 0.1 seconds duration respectively were reported by Graf, Grassl and Aeberhard (1974). Lundgren and Laurell (1987), Graf, Grassl and Aeberhard (1974), and Ahlgren and Owall (1970) recorded occlusal forces in patients during biting and mastication and these are represented graphically in Figure 3.

1.7.3 In-vitro simulation of functional occlusal loading

In-vitro simulation of mandibular movements, tooth contacts and occlusal forces encountered during mastication could enable relatively inexpensive laboratory studies to be used to predict the clinical behaviour of restorative materials. Two problems are; a lack of information on the range of parameters describing the masticatory cycle and difficulties in simulating the complex range of movements and loading conditions encountered clinically.

Gibbs et al. (1981) developed an instrument which used a number of opto-electronic transducers to monitor mandibular movements and reproduce them on a mechanical test rig driven using stepper motors. Although innovative in design this device could not apply a load to the occluding model surfaces.

On the basis of a review of the literature, DeLong and Douglas (1983) considered that an in-vitro load/displacement system capable of simulating masticatory function would need to be able to generate a half sinewave load cycle with an amplitude of 9-180N over a period of 0.25-0.33 seconds and at a rate of 3-4 Hz. These authors developed a two axis servo-hydraulic instrument capable of loading opposing models or extracted teeth under these conditions. DeLong and Douglas (1991) derived the half sine-wave loading cycle used in their instrument from the graphical results of Ahlgren and Owall (1970). Reference to their original work may lead to a different interpretation of the shape of the waveform which may be represented by a half sawtooth (see Figure 3). The loading cycle
Figure 4. Schematic of servo-hydraulic testing machine to simulate mastication (after Delong and Douglas, 1991)
produced using their instrument was generated by two hydraulic actuators, one horizontal and the other vertical. Position and load sensing was monitored in the vertical axis by a load cell and LVDT and used to generate feedback signals. A simplified schematic illustrating this is shown in Figure 4.

It can be deduced from Figure 3 that the loading rate during the masticatory cycle may range from approximately 50-500Nsec\(^{-1}\). The actual period of tooth-to-tooth contact may be relatively short, 100 - 500 msec and functional loads are probably of low amplitude, 10 - 20N. In-vitro simulation of occlusal loads having these parameters is best suited to servo-hydraulic type testing machines which are capable of generating high loading rates and a range of waveforms. The instrument developed by Douglas and DeLong has been used for the measurement of strains in natural and restored teeth (Morin, DeLong and Douglas, 1984).

In-vitro analysis techniques can be applied to determine those areas of teeth which are subjected to high stresses during restoration and subsequent loading. A knowledge of such behaviour can be used to predict how failure may occur clinically. Alternatively the strength of teeth can simply be measured as the force required to cause fracture.

1.7.4 In-vitro measurement of the fracture resistance of natural and restored teeth

In this type of study experimental variables include those related to a load and its application, the type of specimen, its mounting and the restorative procedure.

1.7.4.1 Load application

The characteristics of the loads that have been applied to teeth in-vitro to determine their fracture resistance are quite different from those encountered clinically. The loads produced by a testing machine (often screw driven) operating under stroke control cannot be considered to accurately represent an impact, functional or fatigue type of clinically applied load. The selection of cross-head speeds for such tests appears to have been quite arbitrary and a range of values appear in the literature (see Table 2).

Larson, Douglas and Geistfeld (1981) investigated variations in cavity preparations on the fracture resistance of extracted premolar teeth by applying a controlled load at a rate of 88Nsec\(^{-1}\). Although it was considered to be representative of that observed clinically during mastication it is unlikely that a masticatory load would normally lead to tooth
<table>
<thead>
<tr>
<th>Author</th>
<th>Year of Pubn.</th>
<th>Tooth</th>
<th>Number of Specimens</th>
<th>Cavity Preparation</th>
<th>Restoration</th>
<th>Cross-head Speed</th>
<th>Load Application</th>
<th>Specimen Mounting</th>
<th>Mean Load at Failure (kN)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mondelli et al.</td>
<td>1980</td>
<td>Premolar</td>
<td>10 Sound</td>
<td>0.5mm/min</td>
<td>C.C. Polystyrene</td>
<td>4mm S.B.</td>
<td>1.37</td>
<td>N.G.</td>
<td>0.9</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30 Occlusal</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>N.G.</td>
<td>0.79</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30 MOD</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>N.G.</td>
<td>0.84</td>
</tr>
<tr>
<td>Re &amp; Northing</td>
<td>1980</td>
<td>Molar</td>
<td>N.G. Sound Occlusal</td>
<td>1.0mm/min</td>
<td>5.5mm S.B.</td>
<td>N.G.</td>
<td>2.5</td>
<td>2.4-3.4</td>
<td></td>
</tr>
<tr>
<td>Larson et al.</td>
<td>1981</td>
<td>Maxillary Premolar</td>
<td>12 Sound Occlusal Wide</td>
<td>4.7mm S.B.</td>
<td>Dental Stone</td>
<td>5.2</td>
<td>5.2</td>
<td>2.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>12 Occlusal Narrow</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>3.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>12 MOD Narrow</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>2.1</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>12 MOD Wide</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>3.2</td>
<td></td>
</tr>
<tr>
<td>Share et al.</td>
<td>1982</td>
<td>Maxillary Premolar</td>
<td>10 MOD</td>
<td>0.5mm/min</td>
<td>N.G. Acrylic</td>
<td>1.1</td>
<td>1.1</td>
<td>1.1</td>
<td>1.1</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>10 MOD Unetch Amalgam</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>10 MOD Etch Amalgam</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>1.7</td>
<td>1.7</td>
<td>1.7</td>
<td>1.7</td>
</tr>
<tr>
<td>Bissar et al.</td>
<td>1983</td>
<td>Maxillary Premolar</td>
<td>20 Sound</td>
<td>1.0mm/min</td>
<td>4.8mm wide steel bar</td>
<td>1.1</td>
<td>1.1</td>
<td>1.1</td>
<td>1.1</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>80 MOD</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.7-1.1</td>
<td></td>
</tr>
<tr>
<td>Simonsen &amp; Barouch</td>
<td>1983</td>
<td>Maxillary Premolar</td>
<td>6 MOD</td>
<td>0.02inch/min</td>
<td>N.G. Acrylic</td>
<td>0.83</td>
<td>0.83</td>
<td>0.83</td>
<td>0.83</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6 MOD Unetch Amalgam</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
</tr>
<tr>
<td>Reel &amp; Mitchell</td>
<td>1984</td>
<td>Premolar</td>
<td>10 Sound</td>
<td>0.02inch/min</td>
<td>4.7mm S.B.</td>
<td>Acrylic</td>
<td>1.4</td>
<td>1.4</td>
<td>1.4</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>10 MOD Etch Amalgam</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.79</td>
<td>0.79</td>
<td>0.79</td>
<td>0.79</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>10 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.81</td>
<td>0.81</td>
<td>0.81</td>
<td>0.81</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>10 MOD Etch DBA,Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.74</td>
<td>0.74</td>
<td>0.74</td>
<td>0.74</td>
</tr>
<tr>
<td>Eakle</td>
<td>1986</td>
<td>Maxillary Premolar</td>
<td>16 MOD</td>
<td>0.2inch/min</td>
<td>3.0mm S.B.</td>
<td>Acrylic</td>
<td>1.36</td>
<td>1.36</td>
<td>1.36</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>16 MOD Unetch Amalgam</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.84</td>
<td>0.84</td>
<td>0.84</td>
<td>0.84</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>16 MOD Composite</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>1.02</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>16 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
</tr>
<tr>
<td>Gelb et al.</td>
<td>1986</td>
<td>Maxillary Premolar</td>
<td>7 MOD</td>
<td>0.2inch/min</td>
<td>N.G. N.G.</td>
<td>0.41</td>
<td>0.41</td>
<td>0.41</td>
<td>0.41</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6 MOD Unetch Amalgam</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.41</td>
<td>0.41</td>
<td>0.41</td>
<td>0.41</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6 MOD Composite</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.41</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>14 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>1.2</td>
<td>1.2</td>
<td>1.2</td>
<td>1.2</td>
</tr>
<tr>
<td>Mc Cullock &amp; Smith</td>
<td>1986</td>
<td>Maxillary First Premolar</td>
<td>45 MOD</td>
<td>0.5mm/min</td>
<td>N.G. N.G.</td>
<td>0.41</td>
<td>0.41</td>
<td>0.41</td>
<td>0.41</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>18 MOD Unetch Amalgam</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>1.2</td>
<td>1.2</td>
<td>1.2</td>
<td>1.2</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>9 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>2.23</td>
<td>2.23</td>
<td>2.23</td>
<td>2.23</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>9 MOD Etch DBA,Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>1.83</td>
<td>1.83</td>
<td>1.83</td>
<td>1.83</td>
</tr>
<tr>
<td>Watts</td>
<td>1986</td>
<td>Lower Molar</td>
<td>8 Sound</td>
<td>1mm/min</td>
<td>8mm S.B.</td>
<td>L.C. Resin</td>
<td>4.05</td>
<td>4.05</td>
<td>4.05</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>8 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>2.56</td>
<td>2.56</td>
<td>2.56</td>
<td>2.56</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>9 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>0.31</td>
<td>0.31</td>
<td>0.31</td>
<td>0.31</td>
</tr>
<tr>
<td>Watts et al.</td>
<td>1987</td>
<td>Lower Molar</td>
<td>5 Sound</td>
<td>1cm/min</td>
<td>8mm S.B.</td>
<td>Dental Stone</td>
<td>4.08</td>
<td>4.08</td>
<td>4.08</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>9 Occlusal Prep.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>2.73</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>9 Occlusal Prep.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>2.73</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>8 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>1.47</td>
<td>1.47</td>
<td>1.47</td>
<td>1.47</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>8 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>4.33</td>
<td>4.33</td>
<td>4.33</td>
<td>4.33</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>8 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>2.22</td>
<td>2.22</td>
<td>2.22</td>
<td>2.22</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>8 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>3.3</td>
<td>3.3</td>
<td>3.3</td>
<td>3.3</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>8 MOD Etch Composite</td>
<td>1.5mm/min</td>
<td>N.G. Acrylic</td>
<td>3.16</td>
<td>3.16</td>
<td>3.16</td>
<td>3.16</td>
</tr>
</tbody>
</table>

NG = Not Given
SB = Steel Bearing
DBA = Dentine Bonding Agent
GIC = Glass ionomer Cement

Table 2. Summary of in-vitro loading studies of intact, prepared and restored teeth
fracture.

In a number of studies loads have been transmitted from the crosshead to the specimen by a steel ball or convex metal surface 4.5mm in diameter. Watts, El Mowafy and Grant (1984) varied the position of contacts on specimens using steel balls of different diameters (8.0 and 4.0mm). Joynt et al. (1987) claimed that use of two steel rods to transmit loads allowed optimum adjustment for variations in contour of occlusal surfaces. However, the design of their loading rig would appear to be susceptible to buckling.

The position of occlusal contacts varies between different teeth and, as a result of wear, on the same tooth over a period of time. Figure 5(a) indicates idealised occlusal contacts to be found on a lower first permanent molar in the intercuspal position. Primary contacts are on the marginal ridge and in the central fossa (Shillingburg, Wilson & Morrison, 1984). Contacts also occur between upper and lower teeth in a range of mandibular movements (Figure 5[b]).

Occlusal interferences are atypical contacts which may result in the application of high forces to teeth. Cementation failure, fracture of teeth and of restorations have been attributed to the presence of interferences although little direct clinical evidence supports this (Sheikholeslam & Riise, 1983). An example of potential sites of such interferences on first premolar teeth is shown in Figure 6.

Watts, El Mowafy and Grant (1987) and Sakaguchi et al. (1991) are amongst the few workers who have investigated the effects of variations in load position on the fracture resistance and strains on restored and sound teeth in-vitro. There do not appear to have been any in-vitro studies to measure the fracture resistance or stress distribution of teeth loaded in positions designed to simulate occlusal interferences.

1.7.4.2 Specimen selection and mounting methods

Although it has been reported (Hiatt, 1973) that the lower first molar is the most common tooth to fracture clinically, extracted first premolars have been most frequently used in in-vitro studies, probably because of their wide availability as a result of orthodontic extractions. A number of workers have attempted to reduce the scatter in results which may be attributable to natural variations in size and shape by selecting teeth of comparable size or weight (Eakle, 1986; Joynt et al., 1987; Wieczkowski et al., 1988;
Figure 5. (a) Location of occlusal contacts on a lower first permanent molar
(b) Excursive pathways of opposing cusps on a lower first permanent molar
(after Shillingburg, Wilson & Morrison, 1984)
Watts, El Mowafy & Grant, 1987).

In in-vitro loading studies specimens have often been orientated so that the occlusal plane of the tooth lies parallel to the platen of the testing machine and perpendicular to the direction of load application. This may have been achieved by alignment of the tooth prior to embedding (Watts, El Mowafy & Grant, 1984) or grinding the base of a mounted specimen (Joynt et al., 1987).

A tooth is supported in its socket in-vivo by the periodontal membrane, a complex arrangement of collagen fibres, blood vessels and interstitial fluid, which has an average thickness of 150-300µm (Coolidge, 1937). The mechanical behaviour of the periodontal membrane has been widely investigated but is still not clearly understood. The application of light forces (0.5N) appear to be associated with a displacement of the tooth in its socket which is proportional to the applied load (Picton & Wills, 1978). It has been reported, however, that there may be a logarithmic relationship between load and displacement at low (0.5-2.5N) and moderate (>2.5N) forces (Parfitt, 1960). Recovery of a tooth on unloading can be described in three phases; an initial, rapid, almost linear response, which may be proportional to the deformation of the alveolus. A second phase is associated with recovery of the blood vessels, and a third, small, residual force which restores the tooth to its equilibrium position. Wills et al. (1972) deduced that these phases may be described functionally as three visco-elastic, Voigt type elements in series with decreasing spring constants and/or increasing viscous damping. Korber and Korber (1967) observed a decrease in tooth mobility with an increase in loading rate from 2.5-20Nsec\(^{-1}\). However these loading rates must be considered to be low when compared with the results from Figure 3.

It has been shown that loads applied to teeth result in stresses which are transmitted via the periodontal membrane to the alveolar bone forming the socket walls (Picton et al., 1974). The alveolus is in direct continuity with the trabecular and cortical bone of the mandible or maxilla. The mechanical properties of alveolar bone do not appear to have been studied closely but Wainwright et al. (1976) reported values for Young's modulus of cortical and trabecular bone as 19.0 GNm\(^{-2}\) and 0.3 GNm\(^{-2}\) respectively. Unfortunately the results of the above in-vivo studies were obtained using low loads (0.5-10N) and
Figure 6. Potential sites of occlusal interferences between upper and lower premolar teeth
cannot easily be extrapolated to loading parameters used in in-vitro studies to measure strains and fracture resistance in natural and restored teeth.

Little attempt has been made to accurately simulate the mechanical properties of the supporting tissues in in-vitro measurements of the fracture resistance of teeth. A range of mounting media (chemically and light cured resins and composites, low fusing bismuth/tin alloys and dental stone) and methods have been used in such studies but their mechanical interaction with the specimen has not been studied. It was considered that the mechanical behaviour of extracted teeth mounted in different media warranted further investigation and it was felt that this was well suited to analysis by the finite element method and is the subject of appendix VI.

1.7.4.3 Restorative Procedures

Although fracture resistance has been widely used as a criteria for evaluating materials and procedures used in the restoration of teeth it is a poorly defined term whose precise meaning differs in different studies; Watts (1986), for example, loaded extracted teeth with an increasing force until the point of 'initial tooth fracture', when small fragments were physically detached from the crown. McCullock and Smith (1986) however, applied an increasing load until specimen fractures were complete. Measurements of fracture resistance have been used to investigate the effects of a range of restorative procedures including cavity and endodontic preparations and restoration with a range of materials including amalgam and composites. The question which most of the work in this field has attempted to answer is: Does cavity preparation weaken a tooth and if so, how can a prepared tooth be restored to its original strength? Results from different studies are conflicting and may be attributable to variations in experimental design (specimen selection and numbers) and test methods.

Basic variables associated with cavity preparation include: cavity type, width, depth and the presence of features such as pins, slots and grooves. Unfortunately a lack of standardisation between studies regarding cavity dimensions and instruments used in their preparation makes the interpretation and comparison of results difficult. An example of this is the variation reported by Watts, El Mowafy and Grant (1984) in the strength (2.74 and 4.09 kN) of sound lower molars loaded with steel spheres of different diameters (4.0
Mondelli et al. (1980) observed that preparation of MOD cavities resulted in a lower fracture resistance than teeth prepared with occlusal cavities. In contradiction Larson et al. (1981) observed little difference in strength between these two types of preparation. Loading parameters for a number of such studies are shown in Table 2.

It has been shown that an increase in width of an MOD preparation lowers the fracture resistance of a tooth (Vale 1959, Larson et al., 1981; Blaser et al., 1983). Blaser et al. also reported that an increase in cavity depth further weakened the remaining tooth structure.

Grimaldi and Hood (1973) and Watts (1986) commented on the apparently linear relationship between cuspal deformation and applied load during measurements of fracture resistance. Grimaldi and Hood (1973) however, observed that when teeth were not loaded to failure, elastic recovery took up to ten minutes. This finding is suggestive of non-linear behaviour. Cavity deformation measured as the change in distance between two opposing cavity margins was investigated by Jorgensen, Matono and Shimokobe (1976) and found to be directly proportional to the applied load up to 160N.

In an attempt to reduce the scatter of data attributable to variations in specimen size and shape Mondelli et al. (1974) measured the fracture resistance of cast cobalt-chrome replicas of a lower molar tooth which had been prepared with four different types of cavity. Although statistically, the standard deviation of the results was significantly lower than in studies using natural teeth it is doubtful if the mechanical and fracture properties of such an alloy are comparable with those of enamel and dentine.

The significance of in-vitro measurements of fracture resistance and deformation of prepared teeth is uncertain clinically as teeth are always restored. However, there has been considerable recent interest in comparing the fracture resistance of sound and prepared teeth with those restored with conventional amalgam alloys and etched and bonded composite materials.

A hypothesis has been postulated (DeLong & Douglas, 1983) that an adhesively etched and bonded composite restoration may, together with the remaining tooth, behave as a single structural entity strengthening the prepared tooth against applied loads; whilst a
prepared tooth or one containing an amalgam or unbonded restoration is weaker than a sound tooth. This may be true for a specimen which fails by a load applied axially against the cuspal inclines.

Values for the fracture resistance of restored teeth reported in in-vitro loading studies are conflicting. A number of investigations have shown that teeth restored with amalgam or unbonded and bonded composite restorations exhibit a fracture resistance lower than intact but higher than prepared teeth (Joynt et al., 1987; Wieczkowski et al., 1988; Reel and Mitchell, 1984; Burke, Watts & Wilson, 1990). Other workers have reported that teeth restored with etched and bonded composite materials exhibit a fracture resistance higher than prepared specimens or those containing amalgam or unetched composite restorations (Gelb et al., 1986; Share et al., 1982; Simonsen et al., 1983; Eakle, 1986; Watts, 1986; McCullock & Smith, 1986). Eakle (1986) criticised in-vitro testing of this type for being carried out in an artificial laboratory environment using loading conditions which were unrepresentative of a clinical environment. Stampalia et al. (1986) reported no significant difference in the fracture resistance of extracted premolar teeth restored with MOD amalgam or bonded composite restorations. The authors suggested that a change in slope of the load/displacement curves evident for the composite restorations prior to fracture may have been attributable to adhesive failure at the composite/tooth interface.

In an in-vitro loading study, Morin, DeLong and Douglas (1984) calculated values for the relative stiffness, RS, of teeth restored with bonded composite and amalgam restorations from strain gauge measurements and reported differences in the value of RS between specimens prepared with bevelled and unbevelled cavities and restored with etched composite restorations.

It is evident from the above studies that there is considerable variation in reported values of fracture resistance of prepared and restored teeth (see Table 2.) and it is valuable to see how well in-vitro measurements can be correlated to the clinical behaviour of restored teeth in-vivo.
A comparison of in-vitro test methods with clinical performance of restored teeth

Structural failure of sound and restored teeth may occur as a result of caries, wear (erosion, abrasion or attrition) or fracture. A complete fracture may lead to detachment of part of the tooth or restoration. An incomplete fracture may be manifested clinically as 'cracked tooth syndrome', a term used by a number of workers (Cameron (1964), Stanley (1968) and Braly & Maxwell (1981)) to describe pain related to temperature or pressure on biting which cannot be attributed to other, obvious pathology. In such cases a crack may sometimes be seen clinically.

A number of retrospective clinical studies have attempted to characterise the pattern of tooth or restoration fracture in teeth (Hiatt, 1973; Osbourne et al., 1980; Elderton, 1976; Lemmens et al., 1987; Hansen, 1988). In a retrospective study of endodontically treated premolars restored with MOD amalgam and composite restorations Hansen (1988) observed that 31% of amalgam restorations failed in the first three years compared with none of the teeth restored with composite materials. It should be noted, however, that the composite specimen group was small in comparison to the amalgam (38:168). The pattern of tooth fracture described in this study was most commonly oblique, resulting in the loss of one cusp. In in-vitro testing vertical fractures across the pulpal floor (Stampalia, 1986) have often been seen; possibly as an indirect result of the loading arrangement. In a further study Hansen and Asmussen (1990) calculated the cumulative survival rate for 190 endodontically treated premolar teeth containing MO, DO or MOD composite restorations. The large number of variables (a combination of twenty composite materials and five dentine bonding agents were used) highlighted difficulties in the analysis of a retrospective study of this type. When teeth were categorised according to restoration with a chemical or light curing resin the 5 year cumulative survival rates were 92% and 59% respectively. The authors attributed this surprising result to inadequate polymerisation of the light curing materials.

It is impossible to extrapolate these results to other restorative materials and tooth types. For example, Hiatt (1973) observed that incomplete fractures occurred most
commonly on lower first molar teeth as did bulk amalgam fractures (Lemmens et al. 1987).

1.7.5 Summary of in-vitro and in-vivo methods for determining the fracture resistance of sound and restored teeth.

A number of in-vitro studies have been carried out to measure the strength of restored teeth. In part this work has been prompted by the development of enamel etching techniques, dentine bonding agents and composite materials which, it has been considered, may enable the remaining tooth structure to be strengthened by a restoration. Difficulties in the standardisation of specimen selection, preparation, restoration and testing procedure between different investigations has resulted in wide rangeing and sometimes conflicting results, although there is some evidence that in-vitro the strength of a prepared tooth may be partially restored by placement of an enamel etched composite restoration (Morin, DeLong and Douglas (1984).

Several workers (Hansen, 1988; Eakle, 1986; Wilson, 1990) have questioned the relevance of laboratory test methods to the clinical environment and it is certainly difficult to relate methods used for load application and loading rates (see Table 2.) to in-vivo parameters. It is also likely that the range of test methods used contributes to the variation of results seen in the literature.

There have been a number of retrospective clinical studies of the fracture behaviour of restored teeth (Hansen, 1988) but the large number of variables; type of preparation, restorative materials and technique used, and different operators have made data analysis difficult.

Development of non-destructive in-vitro testing methods to study the mechanical behaviour of sound and restored teeth prior to failure could offer advantages in comparison to measurements of fracture resistance. In-vivo functional loading conditions could be modelled more accurately and as specimens are not tested to destruction a range of restorative procedures may be carried out on the same group of specimens, thereby reducing a large number of indeterminate variables. Experimental stress analysis methods may enable potential failure to be predicted by identification of high stresses and stress concentrations.
CHAPTER 2

MEASUREMENT OF MICRO-HARDNESS AND YOUNG'S MODULUS OF HUMAN ENAMEL AND DENTINE.

2.1 Aims and objectives.

2.2 Method
   2.2.1 Specimen preparation
   2.2.2 Instrumentation
   2.2.3 Test procedure

2.3 Results

2.4 Discussion
   2.4.1 Discussion of the method
   2.4.2 Discussion of the results
2.1 Aims and objectives.

A review of the literature has revealed a lack of knowledge of the micro-mechanical properties of enamel and dentine, specifically Young's modulus and Poisson's ratio. It has been difficult to study variations in these properties throughout a tooth because specimens fabricated for testing using conventional methods have been comprised of a large proportion of the enamel and dentine.

Some success has been reported in the use of micro-hardness measurements to investigate the isotropic behaviour of surface enamel (see Davidson, Hoekstra and Arends, 1974). This technique is considered suitable for measuring variations in hardness with depth for enamel and dentine by indentation of bucco-lingual tooth sections. It has been reported (Marshall, Noma and Evans, 1982) that in addition to hardness, Young's modulus and fracture toughness can also be calculated from measurements taken from Knoop and Vicker's indentations. Lawn and Howes (1981) derived an expression for the recovery in depth of a Vickers impression as a function of the ratio of hardness (H) and modulus (E). This was based on the concept that hardness is an elastic-plastic parameter but unloading is elastic. The equation derived was given as:

\[
\frac{d}{q} = y^2 \cot^2 s - \left[2(1-v^2) y^2 \cot s\right] \frac{H}{E}
\]

where;

\[y = \text{a constant } 0.91\]
\[d = \text{depth of indentation}\]
\[q = \text{contact diameter}\]
\[s = \text{semiangle between opposite pyramidal edges} = \arctan (7/2) = 74.05^\circ\]

The validity of this theoretical analysis was confirmed experimentally from observations on indented surfaces of ceramics. In practice however, this method may be difficult to apply experimentally as the indentation depth, \(d\), is calculated from measurements made by tilting the specimen at different angles in an SEM.

Marshall, Noma and Evans (1982) described a simpler method of calculating the hardness:modulus ratio for a material, based on in-surface measurements of a knoop indentation. The decrease in length of the indentation diagonals caused by elastic recovery
of the specimen was related to the hardness modulus ratio by:

\[
\frac{b'}{a'} = \frac{b - \alpha_1 H}{a - \alpha_1 E}
\]

(2.2)

where \(b/a\) is the ratio of the diagonal dimensions \(a\) and \(b\), in the fully loaded state given by a constant, 0.140646. \(b'/a'\) is the ratio of the altered dimensions and \(\alpha_1\) is a proportionality constant.

It is proposed that this indentation technique is adopted to investigate the elastic properties of specimens of human enamel and dentine at a microscopic level.

2.2 Method

2.2.1 Specimen preparation

Specimens were prepared from permanent lower second molar teeth which had been stored in deionised water at 4°C for a maximum of two weeks following extraction. Teeth were transilluminated and excluded from investigation if there was evidence of cracks, flaws or extraction damage. Three teeth from different patients were selected for preparation.

To facilitate ease of handling and preparation teeth were orientated in individual ABS potting boxes (Type no.; 509-024, RS Components, England [4]) so that the mesiodistal axis of the tooth was parallel to the box walls and the occlusal surface of the crown rested on its base. Specimens were embedded using a chemically polymerising resin (Duralay; Reliance Dental Mfg. Co., USA [5]) and allowed to cure at room temperature for a period of one hour.

Initial preparation entailed sectioning each specimen buccolingually into 2mm slices cut perpendicular to the mesio-distal axis of the tooth. This was carried out using a powered, diamond coated band saw under continuous water cooling at a load of 50g (Exakt; England [6]). Three sections were selected, one from the mid portion of each specimen tooth. Each section was hand ground on a polishing bed (Struers Ltd; Glasgow, Scotland [7]) using silicon carbide papers of 400, 600 and 800 grit progressively under continuous lubrication with water. Care was taken that material was removed evenly from the specimen surface and this was checked periodically by measurement of the specimen
thickness at the margins with a micrometer. Final polishing was achieved by hand lapping on a rotary polishing machine\(^7\) using a 1\(\mu\)m diamond abrasive paste\(^7\). Specimens were intermittently immersed in water to minimise overheating. Each polished surface was examined microscopically at 50x magnification for evidence of surface flaws and if any were present the surface was repolished.

In order to minimise errors due to tilting and avoid the introduction of stresses prior to micro-hardness testing each polished section was mounted on a glass microscope slide and fixed at the margins with a small quantity of cyano-acrylate adhesive (Type No.; 552-804, RS Components [8]). Samples could then be placed firmly on the stage of the hardness tester without rocking.

2.2.2 Instrumentation

Measurements of Knoop micro-hardness were made using a 'Miniload 2' micro-hardness tester fitted with a Knoop indentor (E. Leitz GmbH; Wetzlar, Germany [9]). The instrument was set up in accordance with the manufacturers instructions. The instrument was mounted on foam rubber in an attempt to minimise external vibrations and levelled using the integral spirit level. The hydraulic damping mechanism controlling the descent of the diamond was adjusted for a descent time of fifteen seconds and the diamond positioned so that it was approximately 0.5mm above the specimen surface before release. The microscope objective was aligned so that an indentation was positioned centrally in the field of view.

Instrument calibration was checked by comparing the mean value obtained from five indents of a steel hardness test plate loaded with a test force of 981mN with an DIN standard calibration certificate supplied with the plate and found to be within 1.3%.

2.2.3 Test procedure

Specimens were mounted on the stage of the hardness tester so that there was no perceptible tilt or rock. Orientation was such that the long diagonal of each indentation would lie parallel to the outer tooth surface for enamel and the amelo-dentinal junction for dentine.

A pilot investigation was carried out on an additional specimen to determine the
magnitude of load necessary to produce satisfactory indentations in enamel and dentine. Calculation of Knoop hardness of a material is based on a measurement of the long diagonal of an indentation. However, the method described by Marshall, Noma and Evans (1982) for calculation of the hardness:modulus ratio also required measurement of the short diagonal. An indentation therefore has to be relatively large, with the length of the long diagonal approaching 100μm, to enable the short diagonal to be accurately measured using the optical system in the instrument. It was found that loads of 0.98N for dentine and 4.9N for enamel produced indentations of a size suitable for measurement whilst resulting in minimum surface damage.

Indentation position was planned so that the polished surface of each of the three specimens was indented twenty times in the enamel and twenty in the dentine; The indents in enamel started 300μm from the surface and in the dentine 500μm from the amelo-dentinal junction. Adjacent indentations were spaced by 300μm in enamel and 500μm in dentine to minimise the risk of interactions and crack propagation.

The surface of each specimen was scanned at 10x magnification to identify the position of each indentation. The length of the long and short diagonal was then measured at a magnification of 50x using the vernier scale in the eyepiece of the instrument. If surface damage around the indentation impaired visibility of the apex of a diagonal an estimate was made of its position.

The microhardness tester performed satisfactorily and care was taken to ensure that the Knoop diamond was kept clean and free of grease between indentations. The measurement scale in the eyepiece of the instrument was convenient to use and repeatable measurements with an accuracy of better than 1% could be made when averaged over ten readings. The low refractive index of the moistened specimen surface, especially the enamel, occasionally made the outline of the indentation difficult to see. In these cases, contrast could be improved without affecting measurement accuracy by partially closing the high power focussing aid. On completion of all measurements, specimens were dried, sputter coated with gold and examined under a scanning electron microscope at a magnification of 400x for dentine and 600x for enamel (S8; Hitachi, Japan [10]).

95
2.3 Results

Measurements of the length of the long and short diagonal of each indentation was estimated to 0.1\(\mu\)m, the resolution on the fine scale of the instrument being 0.5\(\mu\)m.

Values of Knoop hardness were calculated for enamel and dentine according to the formula:

\[
HK = \frac{1450F10^3}{b^2}
\]

where \(HK\) = Knoop hardness number

\(F\) = Force in Newtons (0.98N for dentine and 4.9N for enamel)

\(b\) = Length of longer diagonal in \(\mu\)m

By following the manufacturer's recommendations it was estimated that the maximum possible measuring accuracy of the long diagonal was within 2-3\% with a load of greater than 0.147N. This resulted in a 4-9\% possible error in the calculation of hardness.

If it is assumed that there is negligible distortion of the indentor, the ratio of the dimensions of the long and short diagonals (\(b:a\)) of the indentation in the specimen surface when fully loaded will be defined by the geometry of the indentor, \(b:a = 0.140646\). Marshall, Noma and Evans (1982) assumed that the indentation was elastic-plastic in nature but recovery was elastic. Thus, the measured alteration in the ratio of the lengths of long and short diagonals (\(b':a'\)) was considered to be attributable to elastic recovery and could be related to the ratio of the hardness and modulus (see Equation 2.2).

\(\alpha_1\), a proportionality constant, was calculated theoretically by Marshall, Noma and Evans as 1.50 based on an elliptical indentation. However, a value for \(\alpha_1\) of 0.45 was derived from experimental measurements on a range of brittle solids. Measurement variables for hardness and Young's modulus were classified as differences between specimens, between different rows of indentations on the same specimen and at the different indentation distances from the enamel surface for enamel and amelodentinal junction for dentine.

Mean values for hardness and confidence intervals at the 95\% level were calculated for different depths and have been tabulated in Tables 3 and 4 and presented graphically in Figures 7 (a) & (b). A multiple comparison of means (PROC ANOVA; SAS v.6.03)
Table 3. Variation in mean Knoop hardness in dentine with distance from the amelo-dentinal junction

<table>
<thead>
<tr>
<th>Depth (µm)</th>
<th>Knoop Hardness</th>
<th>95% Confidence Interval</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>500</td>
<td>54.5</td>
<td>51.2-57.8</td>
<td>14.1</td>
</tr>
<tr>
<td>1000</td>
<td>61.2</td>
<td>59.4-63.0</td>
<td>9.6</td>
</tr>
<tr>
<td>1500</td>
<td>61.5</td>
<td>59.7-63.3</td>
<td>10.8</td>
</tr>
<tr>
<td>2000</td>
<td>64.3</td>
<td>61.2-67.4</td>
<td>17.5</td>
</tr>
<tr>
<td>2500</td>
<td>62</td>
<td>59.1-64.9</td>
<td>13.7</td>
</tr>
<tr>
<td>Total</td>
<td>60.7</td>
<td>59.4-62.0</td>
<td>17.5</td>
</tr>
</tbody>
</table>

Table 4. Variation in mean Knoop hardness in enamel with distance from the tooth surface

<table>
<thead>
<tr>
<th>Depth (µm)</th>
<th>Knoop Hardness</th>
<th>95% Confidence Interval</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>300</td>
<td>285.7</td>
<td>275.6-295.6</td>
<td>58.4</td>
</tr>
<tr>
<td>600</td>
<td>277.7</td>
<td>268.0-287.4</td>
<td>54.4</td>
</tr>
<tr>
<td>900</td>
<td>277.9</td>
<td>270.5-285.3</td>
<td>42</td>
</tr>
<tr>
<td>1200</td>
<td>262</td>
<td>251.6-272.4</td>
<td>49.7</td>
</tr>
<tr>
<td>1500</td>
<td>257.7</td>
<td>242-273.4</td>
<td>79.3</td>
</tr>
<tr>
<td>Total</td>
<td>272.2</td>
<td>252.9-291.5</td>
<td>79.3</td>
</tr>
</tbody>
</table>
Figure 7. (a) Mean Knoop hardness in enamel with distance from the tooth surface
(b) Mean Knoop hardness in dentine with distance from the amelodentinal junction
Table 5. Variation in mean Young’s modulus in dentine with distance from the amelo-dentinal junction

<table>
<thead>
<tr>
<th>Depth /µm</th>
<th>Young’s modulus /GNm²</th>
<th>95% Confidence Interval</th>
<th>Young’s modulus /GNm²</th>
<th>95% Confidence Interval</th>
<th>Young’s modulus /GNm²</th>
<th>95% Confidence Interval</th>
</tr>
</thead>
<tbody>
<tr>
<td>500</td>
<td>8.7</td>
<td>7.2-10.2</td>
<td>6.5</td>
<td>5.4-7.6</td>
<td>29</td>
<td>24-34</td>
</tr>
<tr>
<td>1000</td>
<td>9.9</td>
<td>8.1-10.7</td>
<td>7.5</td>
<td>6.2-8.8</td>
<td>33.2</td>
<td>27.2-39.2</td>
</tr>
<tr>
<td>1500</td>
<td>10.4</td>
<td>8.5-12.3</td>
<td>7.8</td>
<td>6.4-9.2</td>
<td>34.7</td>
<td>28.2-41.2</td>
</tr>
<tr>
<td>2000</td>
<td>11.4</td>
<td>9.3-13.5</td>
<td>8.6</td>
<td>7.10.2</td>
<td>38.1</td>
<td>30.9-45.3</td>
</tr>
<tr>
<td>2500</td>
<td>11.2</td>
<td>8.6-13.8</td>
<td>8.4</td>
<td>6.5-10.3</td>
<td>37.3</td>
<td>28.6-46</td>
</tr>
<tr>
<td>Total</td>
<td>10.3</td>
<td>9.5-11.1</td>
<td>7.8</td>
<td>7.2-8.4</td>
<td>34.5</td>
<td>31.7-37.3</td>
</tr>
</tbody>
</table>

Table 6. Variation in mean Young’s modulus in enamel with distance from the tooth surface

<table>
<thead>
<tr>
<th>Depth /µm</th>
<th>Young’s modulus /GNm²</th>
<th>95% Confidence Interval</th>
<th>Young’s modulus /GNm²</th>
<th>95% Confidence Interval</th>
<th>Young’s modulus /GNm²</th>
<th>95% Confidence Interval</th>
</tr>
</thead>
<tbody>
<tr>
<td>300</td>
<td>199.7</td>
<td>70.4-329</td>
<td>150.9</td>
<td>53.2-248.6</td>
<td>665.7</td>
<td>234.5-1096.2</td>
</tr>
<tr>
<td>600</td>
<td>551.4</td>
<td>157.4-945.4</td>
<td>416.6</td>
<td>118.8-714.4</td>
<td>1838</td>
<td>524-3152</td>
</tr>
<tr>
<td>900</td>
<td>258.7</td>
<td>-4.3-521.7</td>
<td>195.5</td>
<td>-3.5-394.5</td>
<td>862.5</td>
<td>-15.5-1740.5</td>
</tr>
<tr>
<td>1200</td>
<td>102</td>
<td>69.2-134.8</td>
<td>77</td>
<td>52.2-101.8</td>
<td>340</td>
<td>230.6-1660.5</td>
</tr>
<tr>
<td>1500</td>
<td>290</td>
<td>82-498</td>
<td>219.2</td>
<td>62.1-376.3</td>
<td>967</td>
<td>273.5-1660.5</td>
</tr>
<tr>
<td>Total</td>
<td>299.5</td>
<td>189.7-409.3</td>
<td>226.3</td>
<td>143.4-309.2</td>
<td>998.5</td>
<td>632.4-1364.6</td>
</tr>
</tbody>
</table>
Mean Young's modulus in enamel with distance from the tooth surface

<table>
<thead>
<tr>
<th>Distance from enamel surface (μm)</th>
<th>Young's modulus (GN/m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>300</td>
<td>200 ± 100</td>
</tr>
<tr>
<td>600</td>
<td>800 ± 100</td>
</tr>
<tr>
<td>900</td>
<td>1200 ± 100</td>
</tr>
<tr>
<td>1200</td>
<td>1600 ± 100</td>
</tr>
<tr>
<td>1500</td>
<td>2000 ± 100</td>
</tr>
</tbody>
</table>

95% Confidence Int.

Constant of proportionality = 0.45

Mean Young's modulus in dentine with distance from the amelodentinal junction

<table>
<thead>
<tr>
<th>Distance from amelodentinal junction (μm)</th>
<th>Young's modulus (GN/m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>500</td>
<td>8 ± 4</td>
</tr>
<tr>
<td>1000</td>
<td>12 ± 4</td>
</tr>
<tr>
<td>1500</td>
<td>16 ± 4</td>
</tr>
<tr>
<td>2000</td>
<td>10 ± 4</td>
</tr>
<tr>
<td>2500</td>
<td>6 ± 4</td>
</tr>
</tbody>
</table>

95% Confidence Int.

Constant of proportionality = 0.45

Figure 8. (a) Mean Young's modulus in enamel with distance from the tooth surface (b) Mean Young's modulus in dentine with distance from the amelodentinal junction
was carried out between all groups of variables for values of hardness and indicated a significant difference at the 5% level for dentine between the value 500\(\mu\)m from the amelodentinal junction and all other values. No significant differences were observed for enamel.

Mean values for Young's modulus of enamel and dentine at different indentation distances were calculated by rearranging equation 2.1 in terms of modulus;

\[
E = \frac{\alpha_1 HK}{(b/a-b'/a')}
\] (2.4)

Three values were substituted for \(\alpha\): Marshall's theoretical calculation (1.5), experimentally derived value (0.45) and Omer and Watts (1989) experimental value (0.34). Data is shown together with confidence levels at the 95% level in Tables 5 and 6 and represented graphically in Figures 8 (a) and (b).

Several of the calculated values for Young's modulus of enamel were negative. This occurred when the ratio of lengths of the altered diagonals (\(b'/a'\)) was greater than that of the diagonals under full loading (\(b/a\)). As this was clearly not possible, the negative results which could be traced back to indentations that had sustained surface damage and could have led to a measurement error were excluded from statistical analysis.

As sample groups for enamel did not have equal numbers of specimens a multiple comparison of means based on a generalised linear modelling procedure (PROC GLM; SAS v.6.03) was carried out to examine differences in data between specimens, rows of indentations and distances from the enamel surface or amelo-dentinal junction. No significant differences were found in values of modulus between any groups for enamel or dentine.

A test for normality was applied to both the hardness and modulus data (Shapiro-Wilkes; Univariate procedure, SAS V6.03). Values of hardness and Young's modulus for dentine showed a tendency towards normality but this was less clear for enamel.

2.4 Discussion

2.4.1 Discussion of the method

Unfortunately polishing a specimen surface in order to obtain measurable indentations
Figure 9. Scanning electron micrographs of Knoop indentations in (a) enamel and (b) dentine surfaces
for hardness measurements will alter the surface. It was considered that three factors may influence the hardness measurements when polishing specimens of human enamel and dentine to obtain a flaw free surface of low roughness;

a) Heat generation

b) Differential hardness of enamel and dentine

c) Automated polishing techniques

Generation of heat during polishing was minimised by copious lubrication with water. The differential hardness of enamel and dentine and use of automatic polishing methods may have resulted in preferential removal of the softer dentine. It was for this reason that specimen polishing was carried out by hand using a randomised technique with papers of progressively finer grits and finishing with a 1μm diamond paste(7) on a chamois leather wheel. The thickness of the layer of material removed was noted and in no case was greater than 200μm.

Each of the sections used as specimens were taken from the mid-buccal part of extracted teeth removed from different patients. Following preparation, specimens were stored in deionised water prior to testing to prevent dehydration. A specimen thickness of 2mm was chosen as it has been shown that there may be significant stresses on the surface remote from the indentor if the specimen is too thin (Waters, 1965). The implications of these findings with regard to the indentation of dentine were discussed by Renson and Braden (1971). The minimum specimen thickness recommended by the instrument manufacturers(9) for a Vickers indentation is 10x the depth of penetration of the Vickers diamond; equivalent to 1.5x the length of the indentation diagonal.

The indentor loads selected, 0.98N for dentine and 4.9N for enamel, produced long and short indentation diagonals of approximately 100μm and 14μm respectively. This permitted the short diagonal to be measured with reasonable accuracy. It was evident, however, that surface damage of the enamel occurred during indentation which resulted in cracking and sub-surface chipping (see Figure 9 (a) & (b)). Damage was present to some degree on all the test indentations on enamel carried out with a load of 4.9N and also on trial indentations which were loaded in the range 0.49-9.8N. The significance of this finding is twofold: surface damage made accurate measurement of the length of the short
diagonal for enamel especially difficult. Secondly, fracture of the enamel meant that recovery following indentation was not wholly elastic as energy was used in the formation of the new fractured surfaces. It was considered that this may have influenced calculated values for Young's modulus.

There was no major surface or sub-surface damage evident on the indentations in human dentine and microscopic examination revealed no evidence of cracks radiating from the apices of the indentation. Each indentation covered an area of dentine occupied by a number of tubules (Figure 9 (b)) and it was considered that their presence would not introduce a significant error into the values calculated for Knoop hardness.

Direct measurement techniques using a scanning electron microscope were employed for comparison purposes and possibly improving the accuracy of measurements of indentations in enamel. Difficulties were encountered in levelling the stage to permit measurements to be made perpendicular to the specimen surface without introducing a significant parallax error and this technique was not adopted. It is interesting to note that Lawn and Howes (1981) made use of the parallax shift produced by tilting a specimen to calculate the depth of a Vickers indentation.

In view of work carried out by El Mowafy, Watts and Grant (1989) which established linear temperature coefficients for dentine of -0.014°C⁻¹ for Knoop hardness and -0.073 GNm⁻²°C⁻¹ for Young's modulus, temperature was not included as an experimental variable and all specimens were tested at an ambient temperature of 21°C ± 1°C.

2.4.2 Discussion of the results

Overall mean values for Knoop hardness calculated for human enamel and dentine (Tables 3 & 4) were in good agreement with those reported by other workers (Braden, 1976). Calculation of Knoop hardness is solely dependent on a measurement of the long diagonal, unlike a Vickers indentation where hardness is calculated from the mean of the two indentation diagonals. The range of values and 95% confidence levels suggest that the long diagonal could be measured with reasonable accuracy for both enamel and dentine. Statistically there was no significant difference in Knoop hardness for enamel and dentine between indentations made in different specimens, and different rows. Figure 7 (a) & (b) however, do suggest trends in the variation of hardness from the enamel surface
and amelodentinal junction. There was a decrease in Knoop hardness in enamel with an increase in distance from the enamel surface at a linear rate of approximately -0.023 HK\textmu m^{-1}. Such an observation does not appear to have been reported previously in the literature.

Craig, Peyton and Johnson (1961) observed that the Knoop hardness of dentine was lowest adjacent to the pulp (HK = 30) and amelodentinal junction (HK = 15). Although the precise location of each indentation was not reported and a direct comparison between Craig's work and the results in this study is not possible, the finding that mean hardness was significantly lower for the indentation adjacent to the amelodentinal junction is in good agreement. Mean hardness values for dentine in this investigation (HK = 60.7) were in good agreement with other workers (HK = 50-70; Braden, 1976) but values reported by Craig, Peyton and Johnson (1961) were much lower.

Values for Young's modulus of human enamel and dentine were calculated by application of measurements of the long and short Knoop diagonals to equation (2.4). The proportionality constant in this equation, $\alpha_1$, calculated by Marshall, Noma and Evans (1982) as 1.50 was found to be rather high when compared with the value (0.45) derived from experimental measurements of materials of known modulus. A possible explanation for this is that the theoretical calculation was based on an elliptical indentation. Watts, Omer and Wilson (1989) also derived a value for $\alpha_1$ experimentally as 0.34 based on measurements made on a range of dental composite materials.

It can be seen from Tables 5 & 6 that the calculated value of Young's modulus is heavily dependent on the value of $\alpha_1$ used. It was considered that Marshall's experimentally derived value of 0.45 probably introduced the smallest error. Mean values for Young's modulus of dentine (10.3 GNm$^{-2}$) calculated using a value for $\alpha_1$ of 0.45 were in good agreement with those reported by workers who have used static test methods (5.85 - 19.3 GNm$^{-2}$: Braden, 1976). The 95% confidence levels also suggested that measurement of the short indentation diagonal could be made to a reasonable level of accuracy.

There was no statistically significant difference in Young's modulus between indentations made in different specimens, different rows and distance from the outer
surface or amelodentinal junction. However, results in Table 5 and Figure 8 (a) suggest a trend for dentine that modulus increases with distance from the amelodentinal junction at a rate of approximately $0.0012 \text{ GNm}^2\mu \text{-m}^{-1}$. A variation in Young's modulus throughout the dentine has not been reported by previous workers; probably because preparation methods utilised a large part of the dentine to produce a single specimen. It would seem, therefore that indentation techniques are appropriate for determining the micro-mechanical properties of the dentine.

Calculation of values for Young's modulus of enamel was less satisfactory. Out of a total of sixty indentations in enamel, measurement of the length of the long and short diagonals for each resulted in the calculation of a value for Young's modulus which was negative for twenty specimens. Values for the Knoop hardness of enamel, which depended solely on measurement of the long indentation diagonal, were satisfactory and this suggested possible errors in the measurement of the short diagonal. An alternative, but less probable, explanation is that negative values are the result of anisotropic behaviour of the enamel surface. However, the random distribution of the indentations that produced these results does not support this hypothesis.

Mean values for modulus calculated using a proportionality constant, $\alpha_1$, of 0.45 were relatively high (299.5 GNm$^{-2}$) in comparison to those reported in the literature from which ranged from 24.8 GNm$^{-2}$ (Stanford et al., 1960) to 81.0 GNm$^{-2}$ (Craig, Peyton & Johnson, 1961) when measured using static test methods. This may have been because the fracture process utilises some of the energy that would have been used in elastic recovery resulting in a decrease in in-surface dimensions from the fully loaded condition will not be as great, thus producing a high value for Young's modulus. This finding appears to be in agreement with Lawn and Howes (1981), who stated that measurements of the elastic recovery of a Vickers indentation must be made in the absence of cracking, but is in contradiction to Marshall, Noma and Evans (1982) however, who considered that this method was 'insensitive to cracking and can, therefore, utilise higher load indentations'. The answer may well lie in a comment by Anstis et al. (1981) who advocated the use of indentation techniques only on 'well behaved ceramic materials'.

The experimental value of the proportionality constant, $\alpha_1$, (0.45) derived by Marshall,
Variation of residual Knoop impression dimensions with hardness/modulus ratio

Indentation Dimensions ($b'/a'$)

$0.2$

$0.15$

$0.1$

$0.05$

$0$

$0.02$

$0.03$

$0.04$

$0.05$

$rac{H}{E}$

Alpha $= 0.45$

Enamel - Young's modulus $= 77.9$ GN/m$^2$
(Craig et al. 1961)

Variation of residual Knoop impression dimensions with hardness/modulus ratio

Indentation Dimensions ($b'/a'$)

$0.14$

$0.12$

$0.1$

$0.08$

$0.06$

$0.04$

$0.02$

$0.03$

$0.04$

$0.05$

$0.06$

$rac{H}{E}$

Alpha $= 0.45$

Dentine - Young's modulus $= 12.9$ GN/m$^2$
(Stanford et al. 1961)

Figure 10. (a) Variation of residual Knoop impression dimensions with hardness/modulus ratio for enamel
(b) Variation of residual Knoop impression dimensions with hardness/modulus ratio for dentine
Noma and Evans (1982) was calculated by a regression which was used to fit values of the hardness/modulus ratio plotted against indentation dimensions (b'/a') for a range of brittle materials. Application of such a method was considered in this investigation to derive a value of $\alpha_1$ from the indentation measurements of enamel and dentine by using representative values for Young's modulus of these materials taken from the literature. Figure 10 (a) & (b) show H/E plotted against indentation dimensions (b'/a') for enamel and dentine with the gradient derived by Marshall superimposed on the data. This suggests that the distribution of the data does not follow the slope calculated by Marshall. Such a result is not entirely surprising as it is probably not realistic to attempt to deduce a value for $\alpha$ based on the scatter of measurements for a single material which had a range of H/E of 0.026 (for dentine) in comparison to a range of 0.062 for the materials tested by Marshall.

Use of indentation techniques has shown that values for Knoop hardness of enamel and dentine can be measured accurately and are in good agreement with similar investigations reported in the literature. In addition it has been suggested that the hardness of enamel may vary with depth, a finding which does not appear to have been reported previously. Values for Young's modulus were calculated for enamel and dentine according to the method described by Marshall, Noma and Evans (1982) and the modulus of dentine appeared to vary with distance from the amelodental junction, a result which would have been difficult to establish using conventional macroscopic static test methods.
CHAPTER 3
MEASUREMENT OF THE MECHANICAL PROPERTIES OF AESTHETIC RESTORATIVE MATERIALS

3.1 Aims and objectives

3.2 Method
3.2.1 Specimen materials
3.2.2 Specimen fabrication
3.2.3 Instrumentation
   3.2.3.1 Instrumentation for four-point bend testing
   3.2.3.2 Instrumentation for uni-axial tensile testing
3.2.4 Data collection and storage
3.2.5 Testing procedure
   3.2.5.1 Flexural testing regime
   3.2.5.2 Tensile testing regime

3.3 Results
3.3.1 Flexural testing
3.3.2 Tensile testing

3.4 Discussion
3.4.1 Discussion of the method
   3.4.1.1 Specimen fabrication and design
   3.4.1.2 Flexural test method
   3.4.1.3 Tensile test method
3.4.2 Discussion of the results
   3.4.2.1 Flexural test method
   3.4.2.2 Tensile test method
3.1 Aims and objectives

A knowledge of the mechanical properties of restorative materials is desirable as elastic constants such as Young's modulus and Poisson's ratio are commonly used in calculating stresses from measured strains and in numerical analysis techniques such as the finite element method. A review of the literature has revealed considerable variation in reported values for the physical properties of dental restorative materials which may be accounted for by the different types available, batches variations, mixing techniques and test methods (McCabe et al., 1990). It has not been possible to discover reliable values for the elastic properties of the restorative materials used in this investigation.

In addition to determining values for Young's modulus and Poisson's ratio for these materials it was considered that the sensitivity of the results to variations in test method, cross-head speed and testing temperature warranted investigation. Based on a review of the literature, uniaxial tensile and flexural testing appeared to be appropriate methods for determining the load-displacement characteristics of dental materials. Although diametral compression or 'indirect tensile tests' have also been widely used concern has been expressed about the complex stress distribution and deformation of this type of specimen prior to failure (Zidan et al., 1980). There also seems to be some doubt that it is actually tensile strength that is being measured (Darvell, 1990).

The influence of loading rate or cross-head speed has rarely been investigated but can be of significance when measuring the properties of polymeric materials which may exhibit non-linear behaviour. It was considered, therefore, that elastic properties should be measured at three different cross-head speeds (0.05, 0.1 and 0.2 mm/min.) for tensile and flexural tests. Inter-batch variation may be reduced if the same specimens are used for the different tests and this was considered feasible using low loads below the elastic limit of the material.

The relationship between temperature and the mechanical properties of composite materials has been investigated by Draughn (1981) and did not warrant repetition in this study but it was felt that testing should be performed at two temperatures representative of ambient (≈21°C) and intra-oral conditions (35°C).

Calculation of a value for Young's modulus of a material from a stress-strain plot can
be relatively straightforward but measurement of Poisson's ratio using a quasi-static test method is more difficult requiring simultaneous recording of axial (\(\epsilon_1\)) and transverse (\(\epsilon_2\)) strains on a specimen surface. This could be carried out by measuring strains during a tensile test using dual element 90° strain gauges mounted on the specimen surface.

3.2 Method

3.2.1 Specimen materials

Aesthetic restorative materials were selected for testing which were considered at the time of investigation to represent the current state of the art in commercially available materials. The physical properties of the composite materials were measured in tension and four-point bending and the glass-ionomers in tension alone.

1) A visible light activated, single paste, composite material


2) A chemically activated, two paste, composite material

(P10; 3M Co.). Batch No. P890808 [12]

3) A visible light activated and chemically setting, polyalkenoate lining material

Batch No. P900425 [3]

4) A chemically setting, polyalkenoate lining material

(Baseline; Dentsply, Weybridge, England) [13]

3.2.2 Specimen fabrication

Rectangular beams, 50mm x 5mm x 2mm were fabricated for all groups of materials using a single batch of each material. All materials were stored refrigerated at 4°C and allowed to reach an ambient temperature before mixing. A two part PTFE mould was used to fabricate the composite specimens and it did not prove necessary to use a release agent. The mould was placed on a strip of transparent, polyester film which rested on a microscope slide. The visible light activated composite material\(^{11}\) was extruded into the mould and care was taken to prevent air-bubble entrapment. The mould was slightly overfilled and levelled with a second film and slide in such a way as to allow venting of the excess without causing unidirectional extrusion. Each of the microscope slides cover-
ing the film were, in turn, removed and the material was irradiated by exposure to a visible light source (Visilux 2 Curing Light; 3M Co., Loughborough, England [14]) for a period of 40 seconds. This procedure was carried out on both sides of each specimen and repeated at 5mm intervals along its length. Minimal force was used to extract specimens from the mould on completion of curing and flash that had formed between the film and mould was gently removed with a scalpel. Specimens were then stored at 35 °C. and 100% R.H. for fourteen days prior to testing.

The chemically curing composite material\textsuperscript{(12)} was mixed in accordance with the manufacturer's instructions and in practice it was found that proportions of 3:2, (paste A: paste B) gave a working time of thirty seconds, sufficient to pack the material into the mould. Polymerisation was allowed to proceed undisturbed for five minutes after which specimens were removed from the mould and stored for fourteen days at 35 °C. and 100% R.H. prior to testing.

The visible light activated polyalkenoate cement\textsuperscript{(3)} was mixed in accordance with the manufacturer's recommendations (powder liquid ratio 1:1) using a sufficient quantity of material to fill the mould. The cement was packed into the PTFE mould and a small spatula was used to aid filling and minimise air entrapment. Each specimen was irradiated by exposure to a visible light source for periods of thirty seconds on both sides at intervals of 5mm along its length. Setting and polymerisation was allowed to progress unhindered in the mould for five minutes after which the specimens were removed and any flash removed with a scalpel.

A pilot study was carried out to determine the optimum storage conditions for the polyalkenoate cement specimens. Six glass ionomer\textsuperscript{(3)} samples were fabricated in the manner described above and two of each were stored for a period of twenty four hours at 37°C dry, at 100% R.H. and in an inert silicone oil (Stock No. 556-531; R.S.Components [15]). Specimens were then loaded using PTFE rollers in a three point bending rig at a cross-head speed of 0.1 mm/min. It is generally accepted that the properties of polyalkenoate cements are sensitive to moisture during setting (hence protection of the surface of a glass-ionomer restoration with a varnish following placement clinically). The results of the pilot study indicated that specimens stored dry
became brittle and extremely fragile, while specimens stored wet exhibited visco-elastic behaviour. It was considered that neither of these conditions were representative of the behaviour of the materials in-vivo and it was found that storage in silicone oil appeared to maintain an optimum moisture content which was reflected in the material properties. On this basis specimens of polyalkenoate cements were stored for fourteen days at 37°C in silicone oil prior to testing.

The chemically setting polyalkenoate cement\(^{(13)}\) was mixed in the recommended powder liquid ratio using a quantity of material sufficient to fill the mould. The acrylic acid in this cement is a freeze-dried component of the powder and deionised water was used as the liquid phase. Filling of the mould was performed in the same manner as described previously. The cement was allowed to harden undisturbed for five minutes after which the specimen was removed from the mould, excess flash removed, and stored at 37°C in silicone oil for fourteen days. It was thought that a storage period of fourteen days would enable maximum setting and polymerisation to occur and permit relaxation of any stresses produced in the specimens during fabrication.

All specimens were numbered on removal from the moulds to identify the material type and the sequence of fabrication. Specimens were transilluminated and inspected for evidence of porosity and inhomogeneity. This revealed significant areas of porosity in three specimens, two composite and one glass ionomer cement, which were rejected. Each specimen was measured to an accuracy of 0.01mm in all relevant dimensions prior to testing with three measurements each of thickness, breadth and depth.

3.2.3 Instrumentation

Tests were performed on a Universal Mechanical Testing Machine (Model 1150; Instron, High Wycombe, England \([16]\)) at three different cross-head speeds; 0.05, 0.1, 0.2 mm/min. The machine was allowed to warm up prior to testing to achieve imperceptible zero drift. Electronic calibration was checked frequently and in addition load cell accuracy and linearity was checked using a combination of deadweights\(^{(16)}\) and found to be to an accuracy of better than 1%. Platens, rollers and grips were cleaned of debris after every test.
Figure 11. Four point bending rig for aesthetic restorative materials
3.2.3.1 Instrumentation for four-point bend testing

The four point bending apparatus was based on a design by Swanson (1989) that had been used for study of the mechanical properties of healing experimental fractures (Henry et al., 1968). The apparatus can be seen in Figure 11, with a diagrammatical representation illustrating the testing principles involved in Figure 12. The rig was designed for testing small specimens, with inner loading points being set 20mm apart and outer points adjustable to 40 or 60mm.

Specimen deflection was measured by means of a linear variable differential transformer (Type AG5; Sangamo-Schlumberger, Hastings, England [17]) which measured the movement of one pair of loading points with respect to the other. The transducer was connected to a conditioning amplifier (Type C31)([17]) which produced an output voltage directly proportional to displacement. Calibration of the LVDT was performed by clamping in a jig and measuring the change in output voltage when the core of the transducer was moved by a micrometer screw which had a resolution of 0.001mm (Type M250; Mitutoyo, Japan [18]).

The load measuring system comprised two mild steel cantilever beams to which strain gauges had been attached (see Figure 12.) supporting the lower member of the loading assembly. The four gauges, one mounted on the top and bottom of each beam were connected in a full bridge configuration to a strain gauge amplifier (see appendix III) which produced an output voltage directly proportional to strain. Dead weights were used for calibration and to determine accuracy and repeatability. Care was taken ensure that there was minimal drift of the strain gauge amplifier with time.

Significant features of this design were the use of knife edge supports to reduce the geometrical error present in this type of test. The specimen was supported by freely pivoted roller bearings to reduce friction and the load was transmitted from the testing machine to the cross-head of the rig by means of a hardened steel ball to eliminate misalignment. The four-point bend rig was bolted firmly to the lower platen of the testing machine. No padding or deliberate lubrication of specimens was used for any of the tests.

The apparatus was calibrated and the consistency of stiffness measurements determined by performing tests on mild steel rods of 2.04mm diameter.
Figure 12  Schematic diagram of instrumentation used for flexural testing of restorative materials in four point bending
3.2.3.2 Instrumentation for uni-axial tensile testing

Medium size standard grips\(^{16}\) were fixed rigidly to the lower platen and via a flexible joint to the crosshead of the testing machine. A jig was constructed to enable specimens to be mounted in the grips and clamped by 10mm at each end whilst ensuring parallel alignment to the applied load. Shims of silicon carbide paper (800 grit) were placed between the specimens and the grips to prevent slip. Loads were measured using the integral load cell\(^{16}\) mounted in the cross-head of the testing machine. Calibration was carried out electronically and using dead weight loads, as described above.

Extension was measured using dual element, 90° strain gauges mounted on the top and bottom surfaces of each specimen in the geometrical centre (Type FLA-1AS/11; TML, Japan \([19]\)). Preparation of specimens was carried out by drying and abrading the surface lightly with silicon carbide paper. Strain gauges were orientated and positioned on the centre of each side of the specimen with the aid of a 20x microscope (Weiss Gmbh, Germany \([20]\)) using adhesive tape and bonded in place using a cyano-acrylate adhesive (Type CN; TML, Japan \([21]\)) applied under pressure for a period of three minutes. A small amount of acrylic varnish (Type M-Coat D; Welwyn Strain Measurement, Basingstoke, England \([22]\)) was applied to the grid of each gauge to minimise the risk of electrical short circuit due to moisture.

The orientation of the gauges was such that one element measured the axial strain \(\varepsilon_1\) on the specimen during loading and the other, mounted at 90°, the transverse or Poisson strain \(\varepsilon_2\). The two gauges measuring axial strain, one on each side of the specimen, were connected as a half bridge to a strain gauge amplifier and the transverse gauges were connected in the same configuration to a second amplifier. Calibration of strain measurements was carried out before every test using shunt resistors switched in parallel with one arm of the bridge to simulate a mechanical strain of 1000\(\mu\varepsilon\).

Tensile testing of specimens was performed at two temperatures; 21°C ± 0.5°, and 35°C ± 0.5° by fitting a polycarbonate enclosure around the specimen and grips of the testing machine. Air, warmed by a thermostatically controlled halogen light source was circulated through the enclosure by means of a speed regulated fan. Temperature regulation within 0.5° could be achieved without difficulty. The composite specimens
Figure 13. Schematic diagram of instrumentation used for tensile testing of restorative materials
were tested dry but care was taken that the polyalkenoate cements remained covered with a coating of silicone oil throughout all trials. Temperature was monitored independently by means of a type K thermocouple digital thermometer (Model 51; Fluke Mfg., USA [23]).

3.2.4 Data collection and storage

Data recorded for each specimen prior to testing included material type, batch number, storage conditions and sample dimensions. For flexural testing using the four point bending rig the load cell output, the signal from the LVDT mounted on the test rig and the load output from the strain gauges mounted on the cantilever beams were recorded. This was carried out by connecting the outputs from the conditioning amplifiers to a 16 channel, 12 bit analogue to digital converter fitted to a microcomputer [Figure 12.] (PC26-AD; Amplicon Liveline, Brighton, England [24] & PC-XT type computer; Comcen Technology, Swansea, Wales [25]). A sampling rate of 1Hz was selected for the flexural testing providing the optimum combination for resolution of results and data storage.

Data recorded for the uni-axial tensile testing included the output from the main load cell mounted on the cross-head of the testing machine, and the outputs of axial and transverse strain measured by the gauges. The strain gauge amplifiers were also connected to the A-D converter [see Figure 13.] and data sampled at a rate of 1Hz.

Data was stored for each test run as an ASCII file on floppy disc and subsequently analysed by formatting and calibrating the data using a spreadsheet program (Quattro Pro v2.0; Borland UK, Berkshire, England [26]) and a statistics package (SAS v6.03; SAS, Medmenham, England [27]).

3.2.5 Testing procedure

Following storage, specimens were first subjected to flexural testing then following gauge placement to tensile testing. Composite specimens were carefully dried prior to testing but polyalkenoate cements were kept coated with silicone oil to prevent dehydration.

3.2.5.1 Flexural testing regime

Five specimens for each of the composite materials were in turn positioned on the
Cross-head speed = 0.05mm/min (Specimen 104E)

Figure 14. Load and displacement curves plotted against time for a composite specimen loaded in four point bending
lower rollers of the four-point bending rig. The upper rollers were brought into contact with the specimen until the strain gauge amplifier just began to register a load. The load cell, strain gauge amplifier and displacement transducer were then zeroed and a calibration value of $1000 \mu \varepsilon$ recorded for strain by switching in the shunt resistors. All flexural testing was carried out at an ambient temperature which was monitored at approximately 1 minute intervals using a digital thermometer with the measurement thermocouple placed in close proximity to the specimen. The magnitude of the load to be applied to the specimen had been determined for each type of material in a pilot study and the maximum load was within the linear part of the stress/strain curve. Each specimen was loaded to its maximum and unloaded. This was repeated ten times each at three different cross-head speeds 0.05, 0.1, 0.2 mm/min. A short recovery period of two seconds was allowed between load applications. Following testing, specimens were carefully removed from the rig, examined for evidence of damage or permanent deformation and returned to storage. Any specimen damage observed was recorded and the specimen precluded from further testing.

3.2.5.2 Tensile testing regime

Following flexural testing, strain gauges were bonded to the composite and glass ionomer specimens using the procedure described above. On removal from storage specimens were allowed to equilibrate at the appropriate testing temperature, 23° or 37°C, in the environmental enclosure prior to mounting in the grips of the testing machine. When the temperature had stabilised, the load cell in the testing machine and the strain gauges were zeroed and the gauges calibrated to $1000 \mu \varepsilon$ using shunt resistors. Each specimen was subjected to its maximum load and unloaded. This was repeated ten times for each specimen at three different cross-head speeds 0.05, 0.1, 0.2 mm/min. Again there was a short recovery period of approximately two seconds between subsequent load applications. Load monitoring and logging was as described previously.

3.3 Results

3.3.1 Flexural testing

Logged data stored as an ASCII file for each test run contained four columns,
Figure 15. Variation in displacement with load for composite materials subjected to four point bending at different cross-head speeds.
representing the sample number, load cell output, strain gauge readings and displacement. Each file was calibrated by recalculating the column data using constants derived from the calibration of the LVDT using the micrometer screw and the strain gauges using the deadweight loads. Data at each sampling interval became load in Newtons and displacement in micrometres. An example of the calibrated data is presented graphically in Figure 14.

A data subset produced from the main file contained values of displacement during loading and unloading at 2, 5, 10, 15, and 20N for each specimen at each of the three cross-head speeds 0.05, 0.1, 0.2 mm/min. Mean values were calculated for the displacements at each load for the different cross-head speeds and each material. A linear regression was performed on the load-displacement data for each material at the three cross-head speeds. For each cross-head speed 99.0% of the mean displacements at the different loads could be fitted to a straight line during specimen loading. When the procedure was repeated for the unloading data it was found that the slopes of load against displacement during loading and unloading deviated by less than 2%.

The slope of the load-displacement curves was compared between the different cross-head speeds for the two materials. As specimen groups were relatively small (n=5) it was considered that the application of parametric statistics may not be appropriate and a Friedman, non-parametric analysis of variance(27) was used to compare displacements at each load for both materials at the different cross-head speeds. The result indicated that there was no significant difference in the load-displacement data at any of the cross-head speeds for either material at the 5% level. A graphical representation for both materials of displacement against load labelled with error bars indicating the confidence interval at the 95% level is illustrated in Figure 15. for the different crosshead speeds.

Young's modulus of the two restorative materials and the mild steel reference specimens was calculated from tangents taken along the slope of the load-displacement plots during loading. The tangents were computed at loads between 5 and 15N for the composite specimens and 20 and 100N for the steel specimens. Appropriate values of load and the corresponding displacement were applied to the following formula which had been derived to calculate the deflection of a beam in four point bending at one of the loading
Figure 16. Relevant dimensions for the calculation of the deflection of a beam in four point bending.
Eld = \frac{Wu^2(3l - 4u)}{12} \quad (3.1)

where;

\begin{align*}
E &= \text{Young's modulus} \\
d &= \text{displacement at one of the load points in } \mu\text{m} \\
W &= \text{total force in Newtons} \\
l &= \text{distance between the outer loading points (40mm)} \\
u &= \text{distance between the inner and outer loading points (10mm)} \\
I &= \text{second moment of area:} \\
&= \frac{jh^3}{12} \quad \text{for a rectangular cross section} \quad (3.2) \\
&= \frac{\pi D^4}{64} \quad \text{for a circular cross section} \quad (3.3) \\

D &= \text{specimen diameter}
\end{align*}

The second moment of area \( (I) \) was calculated using equation 3.2 for the rectangular composite specimens and 3.3 for the circular steel specimens by applying the mean dimensions for each specimen.

A mean value for Young's modulus of the steel specimens was calculated as 203.2 GNm\(^{-2}\) with a confidence level at 95% of \( \pm 4.7 \) and this compared favourably with the generally accepted value for mild steel of 207 GNm\(^{-2}\). Mean values for Young's modulus for the composite specimens together with confidence intervals at the 95% level and are presented in Table 7. A non-parametric one-way analysis of variance (Mann-Whitney, PROC NPAR1WAY)\(^{(27)}\) indicated that there was no significant difference in Young's modulus between the two composite materials\(^{(11)(12)}\) at any of the cross-head speeds.

\textbf{3.3.2 Tensile testing}

The calibrated data file consisted of four columns; the sample number, load (N), axial and transverse strain \( (\epsilon_1, \epsilon_2) \). A data subset of the main file contained values of axial and
Table 7. Young's modulus for aesthetic restorative materials loaded in four point bending (GN/m^2)

<table>
<thead>
<tr>
<th>Material</th>
<th>0.05mm/min</th>
<th>0.1mm/min</th>
<th>0.2mm/min</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>mean</td>
<td>C.I.95%</td>
<td>mean</td>
</tr>
<tr>
<td>P10</td>
<td>9.6</td>
<td>8.9-10.3</td>
<td>10.7</td>
</tr>
<tr>
<td>P50</td>
<td>9.5</td>
<td>8.8-10.2</td>
<td>9.5</td>
</tr>
</tbody>
</table>


Figure 17. Axial and tensile strain plotted against load for a range of materials at 23°C and a crosshead speed of 0.1mm/min
Table 8. Young's modulus for aesthetic restorative materials loaded in tension (GN/m^2)

<table>
<thead>
<tr>
<th>Material</th>
<th>Temperature °C</th>
<th>0.05mm/min mean</th>
<th>C.I. 95%</th>
<th>0.1mm/min mean</th>
<th>C.I. 95%</th>
<th>0.2mm/min mean</th>
<th>C.I. 95%</th>
</tr>
</thead>
<tbody>
<tr>
<td>P10 composite</td>
<td>23</td>
<td>13.6</td>
<td>12.5-14.5</td>
<td>12.8</td>
<td>11.9-13.7</td>
<td>13.4</td>
<td>11.7-15.1</td>
</tr>
<tr>
<td></td>
<td>35</td>
<td>12.1</td>
<td>11.5-12.7</td>
<td>12.1</td>
<td>11.0-13.2</td>
<td>12.3</td>
<td>11.0-13.6</td>
</tr>
<tr>
<td>P50 composite</td>
<td>23</td>
<td>10.9</td>
<td>9.3-12.5</td>
<td>11.3</td>
<td>10.2-12.2</td>
<td>12.3</td>
<td>12.0-12.6</td>
</tr>
<tr>
<td></td>
<td>35</td>
<td>9.66</td>
<td>9.1-10.1</td>
<td>9.2</td>
<td>7.5-10.9</td>
<td>10.2</td>
<td>9.3-11.1</td>
</tr>
<tr>
<td>Vitrebond</td>
<td>23</td>
<td>4</td>
<td>3.7-4.3</td>
<td>4.1</td>
<td>3.9-4.3</td>
<td>5.7</td>
<td>2.3-9.1</td>
</tr>
<tr>
<td>G.L.C.</td>
<td>35</td>
<td>3.3</td>
<td>2.8-3.8</td>
<td>3.3</td>
<td>3.1-3.5</td>
<td>3.4</td>
<td>3.1-3.7</td>
</tr>
</tbody>
</table>

Table 9. Poisson's ratio for aesthetic restorative materials loaded in tension

<table>
<thead>
<tr>
<th>Material</th>
<th>Temperature °C</th>
<th>0.05mm/min mean</th>
<th>C.I. 95%</th>
<th>0.1mm/min mean</th>
<th>C.I. 95%</th>
<th>0.2mm/min mean</th>
<th>C.I. 95%</th>
</tr>
</thead>
<tbody>
<tr>
<td>P10 composite</td>
<td>23</td>
<td>0.3</td>
<td>0.25-0.35</td>
<td>0.26</td>
<td>0.23-0.29</td>
<td>0.27</td>
<td>0.23-0.31</td>
</tr>
<tr>
<td></td>
<td>35</td>
<td>0.27</td>
<td>0.24-0.30</td>
<td>0.27</td>
<td>0.21-0.33</td>
<td>0.29</td>
<td>0.24-0.34</td>
</tr>
<tr>
<td>P50 composite</td>
<td>23</td>
<td>0.3</td>
<td>0.26-0.34</td>
<td>0.28</td>
<td>0.24-0.32</td>
<td>0.33</td>
<td>0.23-0.43</td>
</tr>
<tr>
<td></td>
<td>35</td>
<td>0.28</td>
<td>0.26-0.30</td>
<td>0.26</td>
<td>0.22-0.30</td>
<td>0.28</td>
<td>0.25-0.31</td>
</tr>
<tr>
<td>Vitrebond</td>
<td>23</td>
<td>0.36</td>
<td>0.32-0.40</td>
<td>0.45</td>
<td>0.23-0.67</td>
<td>0.37</td>
<td>0.35-0.39</td>
</tr>
<tr>
<td>G.L.C.</td>
<td>35</td>
<td>0.4</td>
<td>0.35-0.45</td>
<td>0.37</td>
<td>0.34-0.40</td>
<td>0.47</td>
<td>0.24-0.70</td>
</tr>
</tbody>
</table>
Figure 18. Axial vs. transverse strain for a range of materials loaded in tension at a cross-head speed of 0.1 mm/min at 23°C
transverse strain during loading and unloading at 2, 5, 10, 15, 20 and 25N for each composite specimen and at 2, 5, 10, and 15N for glass ionomer specimens at three cross-head speeds 0.05, 0.1, 0.2 mm/min and at the two testing temperatures, 23°C and 35°C.

A linear regression was performed for axial and transverse strains plotted against load data for the means of each specimen material at the different cross-head speeds and temperatures. The results indicated that 99.0% of the data for strain at the different loads for all materials at the different cross-head speeds and temperatures could be fitted to straight lines during loading and unloading. It was also evident that the slopes of the load-strain curves during unloading deviated by less than 2% from that during loading. A graphical representation of axial and transverse strain plotted against a range of loads for all materials at a temperature of 23°C and crosshead-speed of 0.1mm/min is illustrated in Figure 17.

Young's modulus for the different materials was calculated from tangents taken along the slopes of the load-extension (axial strain) plots for loads of between 5 and 20N for the composite materials and 5 and 15N for the glass ionomers. Stress (σ) was calculated as the applied load divided by the mean of the original cross-sectional area for each specimen, A₀. Extension was calculated as the measured strain given in microstrain units (ε) divided by the effective length of the strain gauge grid between either end (0.001m). Young's modulus (E) was derived using the following expression:

\[
E = \frac{(\sigma - \sigma_0) / \sigma_0}{(\varepsilon - \varepsilon_0) / \varepsilon_0}
\]  

(3.4)

Mean values and confidence intervals at the 95% level for Young's modulus of each material tested at the different temperatures and cross-head speeds are presented in Table 8. A non-parametric analysis of variance (Mann-Whitney; PROC NPAR1WAY)\(^{(27)}\) was used to compare Young's modulus for the composite\(^{(11)}\) and glass-ionomer\(^{(3)}\) materials at the two different testing temperatures and no significant differences were observed at the 5% level. In addition no significant differences for the values of modulus for each material were observed at the different cross-head speeds at the 5% level by applying a two-way non-parametric analysis of variance (Friedman\(^{(27)}\)).
Poisson's ratio was calculated as the ratio of the transverse to axial strain from data taken on a tangent of the slope of the axial-transverse strain plot at loads between 5 and 20N for the composite materials\(^{(11)(12)}\) and 5 and 15N for the glass ionomer materials\(^{(3)}\).

A linear regression was performed for each material at the three different cross-head speeds and at both testing temperatures during specimen loading and unloading. It was found that 97% of data could be fitted to straight lines for values of axial strains plotted against transverse strains and that differences between the slopes for specimens during loading and unloading were not greater than 3%. A graphical representation of axial strain plotted against transverse strain for all materials at a temperature of 23°C. and at a cross-head speed of 0.1mm/min is illustrated in Figure 18.

Mean values of Poisson's ratio for the three materials tested at the different temperatures and crosshead-speeds are tabulated in Table 9.

Values of Young's modulus derived using the two test methods, flexure and tension were compared for the two composite materials\(^{(11)(12)}\) at the different cross-head speeds and an ambient temperature. This was performed by carrying out a multiple comparison of means (PROC GLM)\(^{(27)}\). The results indicated that there were no significant differences (at the 5% level) for the values of Young's modulus between the different test methods.

### 3.4 Discussion

#### 3.4.1 Discussion of the method

**3.4.1.1 Specimen fabrication and design**

This study was not designed to compare the mechanical properties of a wide range of restorative materials but to provide a range of baseline values for specific elastic coefficients (Young's modulus and Poisson's ratio) that could be used in the derivation of stresses from measured strains. Selected materials were light\(^{(11)}\) and chemically\(^{(12)}\) activated posterior composites and light activated\(^{(3)}\) and chemically setting\(^{(13)}\) glass ionomer cements.

Factors which influenced the design of test specimen included:

a) the depth and area of material that could be thoroughly polymerised by a visible light source,
Specimen Length 30mm
Thickness 2mm
Gauge & adhesive thickness 0.1mm
Gauge construction - epoxy
Gauge Modulus 8.72 GPa
Specimen Modulus 2.5 GPa
2-D Plane Strain Mesh
Element type TPN3

Figure 19. Finite element mesh and contour plot of maximum principal stresses on a tensile composite specimen with bonded strain gauges attached.
b) the minimum dimensions necessary for placement of strain gauges for the tensile testing,

c) the dimensions of the four point bending rig,

d) the design of specimens used by other workers.

It was considered that the maximum satisfactory curing depth and diameter for a posterior composite (universal shade) using a visible light source\(^{(14)}\) was 2 and 5mm respectively and this defined the depth and width of the specimen.

A rectangular specimen of dimensions 25 ± 2mm x 2 ± 0.1mm x 2mm ± 0.1mm is advocated in British Standard BS5199 (1989) for flexural testing of resin based filling materials. Chitchumnong et al. (1989) compared the flexural properties of dental polymers in three and four point bending using specimens with a span-to-depth ratio of 20:1 and recommended a minimum ratio of 16:1. This appears to be significantly higher than the equivalent ratio taken from BS5199 of 12.5:1.

The rectangular beam of dimensions 50mm x 5mm x 2mm used in this investigation fitted satisfactorily in the four point bending rig, spanning the lower outer rollers which were set 40mm apart. A specimen of these dimensions was comparable to those used in similar studies by;

- Oysaed and Ruyter (1986) [25 x 2 x 2 mm],
- Braem et al. (1989) [35 x 5 x 1.5mm],
- Bryant and Mahler (1986) [12 x 4 x 1mm],
- Ban and Anusavice (1990) [36 x 3.1 x 3.1mm]

The axial and transverse strains recorded by the gauges in the tensile test should be representative of the bi-axial strain field in the specimen. The gauges selected had a grid area of 1mm\(^2\) which was considered to be small enough not to be influenced by edge effects. In order to ensure that the strain field on the specimen would be uniform in the region of gauge placement however, a two-dimensional plane strain finite element analysis was carried out on the specimen. Appropriate dimensions, load case and material properties were assigned to the model (Figure 19). The results indicated that the stress field in the area of gauge placement was uniform and representative of that over a large part of the specimen.
3.4.1.2 Flexural test method

On the basis of the findings of a number of workers (Chitchumnon, Chung, 1989; Chung, 1990) and an analysis of potential errors in the bend test (Swanson 1989) it was considered that three point flexural testing was not appropriate for the investigation of the elastic properties of restorative materials. A four point bending rig had been designed by Swanson based on the principles he had discussed. As in standard test rigs the specimen was supported by freely pivoted roller supports but in this case errors due to geometry were minimised by pivoting the supports about an axis parallel to the roller’s axis and displaced by a distance equal to the radius of the roller (see Figure 12.). If the pivot points are only able to move vertically, the separation of the points of contact with the specimen horizontally will be unaffected by its deflection. Although it is desirable that deflection is measured independently in the centre of the specimen, the rig was designed to test small specimens and deflection was measured, using an LVDT, as the change in distance between the two pairs of loading points.

The rig performed satisfactorily and it proved straightforward to position specimens and apply a load. It was not found necessary to use a lubricant to reduce friction between the specimen and the rollers although the silicone oil used to prevent dehydration of the polyalkenoate cement specimens may well have had a lubricating action. A number of workers (Darvell, 1990) have advocated the use of padding such as a polythene film to improve the uniformity of the load distribution during flexural testing of brittle materials. Such padding was not found to be necessary as the pivot supporting the lower rollers helped to ensure a uniform load distribution. During testing the crosshead was lowered until the strain gauge output just registered a load. Specimen loading and unloading was then performed at the designated cross-head speed and up to the maximum load. This was carried out by manual switching of the cross-head controls and resulted in very slight variations in zero and maximum loads. It was for this reason that values of Young’s modulus were calculated from a tangent of the load/displacement plot.

3.4.1.3 Tensile test method

The elastic behaviour of the aesthetic restorative materials used in this investigation
was studied in both flexure and uni-axial tension. The standard method for observation of the mechanical properties of a material in tension entails measurement of the extension of a specimen subjected to a range of known loads, often transmitted via the grips of a testing machine. Load can usually be measured satisfactorily by means of a load cell mounted on the testing machine and extension may be measured in a number of ways including the use of extensometers, optical transducers or the amount of separation between the specimen grips. The small size of specimens in this study would have made the first two of these difficult to perform and measuring the amount of grip separation may be an inaccurate method because a small movement between the specimen and the grip would constitute a large error in the overall measurement of extension. Use of electrical resistance strain gauges overcame this problem enabling the direct measurement of both axial and transverse strains on the surface of the specimen.

Values for Young’s modulus of the materials were calculated from the load cell and axial gauge data and Poisson’s ratio was derived as the ratio of transverse to axial strains. Use of gauges on either side of the specimen produced a slight increase in gain and helped to compensate for eccentric loading although a jig was used to ensure specimens were mounted axially.

The restorative materials tested were poor thermal conductors and did not effectively sink heat generated by the strain gauges. Although it was only possible to use a very low supply voltage (0.5V) to prevent the gauges overheating satisfactory results were obtained (see appendix II).

It is important that thermal properties are closely matched between a strain gauge and the substrate material. Fluctuations in environmental temperature or the heating effect produced by passing an electric current through a gauge may result in differential thermal expansion between the gauge and substrate and lead to an incorrect measurement of strain. The value for the coefficient of thermal expansion of composite restorative materials has been reported to be approximately $10^{10}\text{oC}^{-1}$ (Craig, 1989) and gauges were selected which matched this as closely as possible having a coefficient of thermal expansion of $10^{11}\text{oC}^{-1}$.

It has been suggested that strain gauges may reinforce specimens fabricated from low
Figure 20. Strains measured on the surface of a tensile composite specimen having a range of values of Young's modulus and subjected to a load of 25N - derived using the finite element method.

- Measured strain on specimen surface
- Measured strain on strain gauge surface

Modulus of gauge & adhesive = 8.7 GPa
modulus materials [<10GNm$^{-2}$] (Perry 1990, Little et al., 1990) resulting in incorrect values of strain. It was considered that this phenomenon should be investigated further as some of the materials, particularly the polyalkenoate cements\(^{(3)}\), investigated in this study may have low values for Young's modulus. A two-dimensional plane strain finite element analysis was performed on a scale model of a cross-section of the tensile test specimen and strain gauges used in this investigation. An automatic technique was used to generate the finite element mesh resulting in slight irregularities although these did not have affect the accuracy of the solution. An axial load of 25N was applied to the specimen which had been restrained in the X and Y direction along its lower border. A range of values for Young's modulus and a value of Poisson's ratio of 0.3 were assigned to the model. The resultant strain was measured between two nodes at the midpoint on the surface of the specimen for the different values of E. The model was then modified to include two strain gauges bonded to the specimen surface (see Figure 19.). Appropriate values for the elastic constants of the gauge material and adhesive (Scott, 1991) were assigned to the model and the analysis repeated. The results indicated (Figure 20.) an increase in the difference between the strains measured directly on the specimen and those measured on the gauge with a decrease in the modulus of the specimen. This may be attributed to the relative stiffness of the gauge which reinforced the specimen and resulting in a possible error of 10% in a measurement of strain if the specimen had a modulus of 5 GNM$^{-2}$ or less.

3.4.2 Discussion of the results

The aim of this part of the investigation was to determine the mechanical behaviour of representative types of restorative materials subjected to loading and establish values for the principal elastic coefficients namely Young's modulus and Poisson's ratio. Experimental variables which may influence these values were considered to be the test method, cross-head speed and temperature. A number of studies have investigated these factors individually (Draughn, 1981; Chitchummong, Brooks & Stafford, 1989) but their effect in combination has not been widely investigated. The testing temperatures used in this investigation did not represent extreme conditions, but it was considered that 23°C and 35°C spanned the range of temperatures to which these materials are most commonly
exposed in-vivo. A wide range of cross-head speeds have been reported in the literature often without comment. High speeds (1.0-2.0mm/min) may be suitable for measuring the fracture strength of specimens but can make calculation of Young’s modulus from a secant or tangent of the stress/strain plot difficult. Relatively low cross-head speeds were therefore used in this study (0.05, 0.1 and 0.2mm/min).

It was hoped that a chemically curing polyalkenoate cement\(^{(13)}\) could have been tested but a pilot study revealed that this type of material was exceedingly brittle and failed in flexure and tension at load of less than 15N. The light activated polyalkenoate cement\(^{(3)}\) was also relatively brittle and it proved difficult to mount specimens in the four-point bending rig without fracture.

### 3.4.2.1. Flexural test method

A pilot study revealed that the two composite materials failed in bending by brittle fracture at a mean load of 54.5N (n=5). It was considered therefore that loads in the range from 0-25N would be appropriate to determine the elastic properties of these materials.

The periodic spikes observed on the plot of load in Figure 14. could have been caused by slight changes in the seating of the rollers on the specimen but as they were not evident on the chart output from the testing machine are more likely to be artefactual, and due to electrical interference.

It was found that Young’s modulus was easier to calculate from a tangent of the load displacement plots than as a secant because manual control of the cross-head movement resulted in slight variations in the position of the zero and maximum load which could have influenced the resultant slope significantly.

The mean value for Young’s modulus of the steel reference specimens calculated using equation 3.1 (203.2 GNm\(^{-2}\)) compared favourably with that reported generally. It was therefore slightly surprising, that the mean values for Young’s modulus of the restorative materials were so low (see Table 7.). Values for the composite materials\(^{(11),(12)}\) were closely comparable at all cross-head speeds and confidence levels at 95\% were generally less than 10\% of the mean which suggested, especially with relatively small sample numbers, an acceptable degree of error. It is possible that the comparable value of
modulus for the two materials may be accounted for by the higher filler content of one\(^{(11)}\) relative to the other\(^{(12)}\) and the higher polymer conversion ratio of the chemically cured material\(^{(12)}\) relative to the light cured composite\(^{(11)}\). Values of modulus determined in this study appear to be rather lower than those for comparable materials tested under similar conditions and reported elsewhere in the literature. Bryant and Mahler (1986), for example, measured values for the modulus of elasticity of 18.8 and 16.9 G\(\text{N}\cdot\text{m}^{-2}\) for two chemically activated composites\(^{(12)(28)}\) when subjected to three point bending. This difference may in part be explained by the use of a high cross-head speed (1.25 mm/min) and the three point loading configuration. Chitchumrong, Brooks & Stafford (1989) measured the elastic modulus of a range of unfilled denture base polymers in four-point bending and reported a range of values from 1.4 to 2.4 G\(\text{N}\cdot\text{m}^{-2}\). These values provide a degree of correlation with the results in this study as it is acknowledged that the introduction of large quantities of high modulus filler particles (typically 80-85\% by weight) significantly increase the stiffness of these materials.

Loading tests were performed on each specimen at three different cross-head speeds 0.05, 0.1 and 0.2 mm/min as it was thought that this may influence the elastic modulus of the materials. These values of crosshead-speed enabled accurate manual control of the testing machine without overshooting the designated loads. A non-parametric analysis of variance revealed that there were no significant differences in the values of modulus for the two materials at the chosen speeds. In addition, examination of the slopes of the load-displacement plots (Figure 15.) does not suggest the presence of any trends.

### 3.4.2.2 Tensile test method

A pilot study revealed that polyalkenoate cement specimens failed in tension by fracture at a mean load of 31.2 N (n=5). A loading range of 0-15 N for the glass-ionomer and 0-25 N for the composite specimens was therefore adopted. A regression applied to the load-strain data for the composite and light curing glass-ionomer materials at the different cross-head speeds established that there was a linear relationship between load and strain for the different materials and test variables within the specified loading range. The slopes of the loading and unloading regression lines differed by less than 2\% and this
Figure 21. Variation in peak axial strain with time for a polyalkenoate specimen loaded in tension.
indicated that the composite materials[11](12) and the polyalkenoate cement[3] exhibited linear elastic behaviour when loaded in tension at the specified cross-head speeds from 0-25N and 0-15N respectively.

During the testing period for each specimen (approximately thirty minutes) it was observed that the light curing glass ionomer samples[3] exhibited non-linear behaviour which manifested itself as a decrease in the peak strain attained when specimens were loaded to 15N. When peak axial strain was plotted against loading duration for the glass ionomer specimens[3] (Figure 21.) the peak strain decreased exponentially indicating a significant time dependent deformation and characteristic visco-elastic behaviour. Although the glass ionomer material was linearly elastic during each loading cycle the long term visco-elastic specimen behaviour necessitated recalibration of strains at the commencement of each loading cycle rather than simply at the commencement of the testing regime for each specimen. Overall this did not significantly influence the findings in the study but may be of significance clinically if such materials are used to line cavities and play a role supporting an overlying composite or amalgam restoration when there is a possibility they may undergo a long term deformation.

Values for Young's modulus were again calculated from tangents rather than secants of the plot of load against axial strain. There were significant differences in the values of elastic modulus between the glass ionomer[3] and composite materials[11](12), the glass ionomer having low values for Young's modulus in the range 4-5.7GNm⁻². This is not entirely surprising as the material has a relatively high water content. The low elastic modulus of this material may be significant in relation to the measuring technique as a numerical stress analysis (see discussion earlier) suggested that strain gauges may cause local specimen reinforcement when mounted on materials having a modulus lower than 7GNm⁻². There was no significant difference in the values of Young’s modulus between the two composite materials[11](12) although trends suggested that the chemically cured material had a slightly higher modulus which may be related to a higher polymer conversion ratio. A significant decrease was apparent in values of modulus for the light curing composite and glass ionomer material at the higher testing temperature (35°C). As in the flexural test method, cross-head speed did not seem to have any significant effect
although a slight increase in modulus could be observed for the light curing composite\(^{(11)}\) and glass ionomer cement\(^{(3)}\) at higher speeds.

Mean values for Poisson's ratio were calculated as the ratio of the transverse to axial strain. An example plot in Figure 18 shows the high axial strain exhibited by the glass ionomer material\(^{(3)}\). This graph also illustrates the linearity of the relationship of axial and transverse strain over the load ranges. Values of Poisson's ratio calculated in Table * were considered to be satisfactory. There is little evidence in the literature of quasi-static measurements of Poisson's ratio for aesthetic restorative materials and reported values using dynamic measurements (Whiting & Jacobsen, 1980[a]) for comparable materials are slightly lower than those reported here (0.27)\(^{(29)}\). Confidence levels were within approximately 10% of the mean values, with one exception and this was considered to be very satisfactory especially in view of the extremely small actual displacements measured by the transverse strain gauges (0.01-0.1\(\mu\)m). Significant differences in values of Poisson's ratio were not apparent between the different materials and at the different cross-head speeds and testing temperatures. However, trends indicated that the glass ionomer material had a higher Poisson's ratio than the composites and this may be accounted for by the high water content and absence of filler.
CHAPTER 4

IN-VITRO FULL FIELD STRESS PATTERN ANALYSIS OF INTACT
AND RESTORED TEETH BY MEASUREMENT OF THERMOELASTIC EMISSION

4.1 Aims and objectives

4.2 Method
  4.2.1 Specimen preparation
    4.2.1.1 Cavity preparation
    4.2.1.2 Cavity restoration
  4.2.2 Instrumentation
  4.2.3 Test procedure
  4.2.4 Calibration procedure

4.3 Results

4.4 Discussion of the method and results
4.1 Aims and objectives

One of the aims of this investigation was to determine the magnitude and direction of the principal stresses on the surface of teeth subjected to a range of loading conditions. Strain gauges are able to resolve all the data necessary to fully describe a strain field between two points on a specimen surface. Although electrical resistance strain gauges may be small (<1mm²) the complex geometry and small size of natural, human teeth makes the application of multiple gauges on a tooth surface impractical. It is therefore desirable to be able to position a gauge on a region of high strain where potentially, failure may occur. A full field analysis technique, such as brittle coatings, can be used to identify such a region but these too have disadvantages when applied to teeth. Brittle coatings fail by cracking in tension, at a threshold strain which is often high. Photoelastic models composed of more than one material can be difficult to fabricate and it may not be easy to relate the results to an actual specimen.

The principles behind a non-contacting full-field stress analysis technique based on the thermo-elastic phenomenon has already been described (see Section 1.3.2.3). It was considered that this could be used to investigate the distribution of stresses over the surface of prepared and restored teeth subjected to dynamic loads and it was anticipated that the results could be used to aid subsequent strain gauge placement.

4.2 Method

4.2.1 Specimen preparation

Lower second permanent molar teeth were selected as specimens and chosen for having a standard molar crown form and average dimensions (bucco-lingual width, 10mm; mesio-distal width, 10.5mm; occluso-cervical crown height, 7.0mm; Wheeler, 1971). Specimens were examined radiographically and transilluminated; any exhibiting cracks, evidence of extraction damage or caries were discarded. Specimens were stored in deionised water and refrigerated at 4°C for a maximum period of two weeks prior to testing. Each tooth was positioned in a brass ring with internal dimensions of 15.5mm diameter and 18mm long (Stock No. 619-985; R.S. Components, Corby, England [30]) so that the cervical margin was 1mm above the top of the ring and the occlusal plane of the tooth was perpendicular to its long axis. This procedure was aided by use of a silicone
matrix. Teeth were then embedded to the top of the ring using a highly filled chemically curing acrylic resin (Formatray; Kerr, Romulus, USA [31]) mixed in the proportions, three grams of powder to one millilitre of liquid. Mounted specimens were numbered and stored under water at 4°C until ready for testing.

4.2.1.1 Cavity preparation

As part of the testing procedure mesio-occluso-distal (MOD) cavities were prepared in all the specimens. This was performed with a standardised technique using a pear shaped tungsten carbide bur (331L; Beavers Dental Co., Ontario, Canada [32]) in a high speed dental handpiece under constant lubrication. Cavity sizes were kept as constant as possible and mean dimensions were; 3.5mm wide buccolingually and 4.1mm high occluso-cervically. Cavities were not prepared with mesial or distal boxes and there was no narrowing at the isthmus. Dimensions were chosen so that the width was approximately one third the width of the tooth and a tripod occlusal contact could be produced by resting a 4.5mm steel ball bearing on the cuspal inclines. This was designed to simulate an intercuspal occlusal contact similar to that seen in-vivo. The testing procedure was performed prior to, and immediately following cavity preparation and the cavity was kept moist throughout the procedure with a dampened cotton wick.

4.2.1.2 Cavity restoration

Following preparation and testing specimen cavities were restored in two ways; firstly with an amalgam and then with an enamel etched and dentine bonded composite restoration. Each cavity was dried thoroughly and a flexible metal matrix band fitted around the tooth (Automatrix; L.D. Caulk, U.S.A [33]). The band was burnished to obtain good adaptation at the mesial and distal margins. A dispersed phase amalgam alloy (Dispersalloy; Johnson & Johnson, USA [34]) was triturated and packed into each cavity using a range of hand instruments. The material was condensed thoroughly to ensure that voids were kept to a minimum. Occlusal anatomy was carved on the surface of the restoration while the amalgam was still relatively soft. The aims of this were to restore the original tooth contour as closely as possible and also ensure that the steel ball bearing used for load application remained clear of the restoration, only contacting the remaining
tooth structure. Amalgam restorations were allowed to harden prior to testing for 24 hours at a temperature of 36°C and 100% relative humidity by placement on a tray in a temperature controlled water bath (Model 160; Grant Ltd., Cambridge, England [35]).

The amalgam restorations were removed from each specimen following testing. This was carried out using a pear shaped tungsten carbide bur in a high speed dental handpiece with lubrication to carefully dissect out the amalgam restoration whilst removing minimal further tooth structure. Prior to placement of the composite restorations the enamel at the cavity margins was etched for 30 seconds by application of a gel containing 36% phosphoric acid (3M (UK), Loughborough, England [36]). Each cavity was washed and dried thoroughly for a period of 20 seconds using a triple syringe following the etching procedure. A dentine adhesion promoter (Scotchbond 2; 3M (UK) Batch No. P900325 [37]) was applied to the dentine surfaces in accordance with the manufacturer’s recommendations; a priming agent containing maleic acid and hydroxy-ethyl-methacrylate (HEMA), was applied to the prepared dentine surface and the surface dried after 20 seconds. A light curing adhesive containing HEMA and bis-GMA was then applied and exposed to a visible light source\(^{14}\) for 20 seconds. The prepared cavities were restored with a light curing dental composite material\(^{11}\) considered suitable for use in posterior teeth. A polyacetate matrix band\(^{33}\) was placed around each specimen prior to restoration. The composite material was placed in each cavity in three equal, horizontal increments approximately 1.2mm thick using hand instruments and packed thoroughly against the cavity walls to ensure good adaptation and minimise air entrapment. Each increment was polymerised by exposure to a visible light source\(^{14}\) for a period of forty seconds. The occlusal surface of the final increment was carved to restore the original anatomy of the tooth and again to ensure that the steel bearing used for loading did not contact the restoration. Specimens restored with composite materials were stored for a minimum of two hours at 36°C. and 100% R.H. prior to testing.

4.2.2 Instrumentation

Specimens were mounted in a loading rig which enabled mesio-distal rotation, tilting buccolingually and translation in the mesio-distal and bucco-lingual directions. This range
Figure 22. Schematic diagram of instrumentation used for thermoelastic stress analysis of teeth
of adjustment enabled precise positioning of loads transmitted via a steel ball bearing or tapered rod to the surface of the crown of the tooth. The design and construction of the rig is discussed in further detail later (section 5.2.5.2). The mounting plate of the testing rig was bolted rigidly to the lower platen of a uni-axial servo-hydraulic testing machine (Type 9500; Dartec, Worcester, England [38]). The testing machine had a 20KN load frame and was fitted with a 20KN tension/compression load cell and 2KN subcell. Loading parameters could be programmed in terms of amplitude, frequency, and waveform and the machine was capable of full load and stroke control. For the purposes of the thermoelastic analyses specimens were orientated so that the occlusal surface was perpendicular to the direction of load application and an axial load was applied via a steel bearing (4.5mm diameter) which produced a tripod occlusal contact on the tooth surface. Loading position was checked carefully by marking areas of contact between the ball bearing and tooth with an articulating foil[39]. The testing machine was programmed to apply a dynamic, compressive load with a sine waveform and at an amplitude of 200N and a frequency of 20Hz. This was considered to be well above the 10Hz threshold for quasi-adiabatic behaviour and there was minimal signal loss due to thermal diffusion through the specimen. Although a load of 200N is probably greater than that normally encountered in function clinically it provided the optimum sensitivity for the SPATE scanner. Electrical damping of the feedback control from the load cell to the hydraulic servo-valve was adjusted by altering the gain of the load cell amplifier to minimise vibration. The subcell output was connected to the analogue controller in the testing machine.

The SPATE system used to perform the thermoelastic stress measurements required a trigger signal from the load cell of the testing machine which was fed into the phase correlator forming a phase locked loop (Figure 22.). The tripod mounted SPATE scanner head was positioned so that its viewing port was in direct line of sight with and 0.75m away from the specimen. This gave a spatial resolution of 1.0mm and a field of view of 25°.

The infra-red detector was scanned across the specimen surface in a raster like manner using a series of stepper motors to give a full-field graphical representation of the stress
distribution. This could be displayed in real time and the data stored on floppy disc for later analysis. Unfortunately it was not possible to export the data in a form which could be analysed independently of the SPATE software.

4.2.3 Test procedure

Prior to testing, the surface of the crown of each specimen was coated with an even layer of matt black paint (Stock No. 567-345; R.S. Components, Corby, England [40]) which was allowed to dry thoroughly. The purpose of this was to make the specimen behave as a black body radiator by raising its emissivity and improving the thermal transfer characteristics. Further coats were not necessary following the restorative procedures as the surface did not degrade. The coating chosen was that recommended by Harwood and Cummings (1986) which had a high emissivity (0.93) and did not cause any significant signal attenuation due to thermal lag at test frequencies of 10Hz or greater. It was estimated that the thickness of the paint on the enamel surface was not greater than 100μm which according to Belgen (1968) would result in a signal attenuation of less than 5% and this was confirmed by the experimental results.

Initially unprepared specimens were mounted in the loading rig and rotated mesio-distally so that the buccal surface faced the SPATE scanner head. A dynamic load, having characteristics as described above, was applied to the occlusal surface of the tooth surface via the steel bearing. The limits of the scanning area on the buccal surface were defined by tracking a visible light cross-hair around the borders of the target area. The surface was scanned at a rate of 1 measurement per second at a spatial resolution of 1.0mm and an overlap of 75%. This resulted in a scanning period of approximately 3.5 minutes. The procedure was repeated for the lingual surface following rotation of the specimen. The mesio-distal orientation of the buccal and lingual surfaces of each specimen was recorded to ensure that specimens could be precisely repositioned for testing following the various restorative procedures. Scans were repeated for the buccal and lingual surfaces of both specimens following cavity preparation and restoration with amalgam and etched and bonded composite materials. During and between the scanning procedures specimens were maintained at an ambient temperature of 23°C. Specimens were examined at
frequent intervals for evidence of structural damage due to loading. It was not possible to transilluminate the teeth but the coating was inspected for cracking. On completion of the test procedure scans were plotted and analysed for areas of high stress.

4.2.4 Calibration procedure

In order to be able to quantify the results from the thermoelastic analysis in terms of principal stresses it was necessary to apply a value for the thermoelastic constant \( (K_m) \) of the test materials. This constant may be calculated theoretically (see 1.3.2.3 earlier) or derived experimentally. A value for the thermoelastic constant of human enamel has not been determined and it was decided that a theoretical derivation should be supported by experimental measurements. This was performed by comparing the sum of the principal stresses in a region of uniform stress on a specimen calculated from strain gauge measurements with a SPATE plot calibrated using a theoretical value for \( K_m \).

As part of a pilot study an unprepared natural tooth was mounted in the test rig and subjected to the loading parameters described earlier. The buccal surface of the specimen was scanned using the SPATE system and an uncalibrated plot of the stress distribution produced. An area of the specimen surface was identified on the plot which exhibited a uniform stress distribution and the matt black coating was removed from this region of the specimen surface using a solvent. A three element rosette strain gauge (Type FRA-1AS/11; TML, Japan [41]) was bonded to the enamel using a cyano-acrylate adhesive (for a more detailed description of this procedure see 5.2.1) and connected via a three channel bridge amplifier to an analogue-to-digital converter\(^{24} \) fitted in a micro-computer\(^{25} \). Data was logged at a sampling rate of 1KHz during the loading procedure. The magnitude and direction of the principal stress were calculated from the calibrated, individual element readings (see Section 5.3.1 and Appendix IV) using a mean value for Young's modulus taken from the literature \( (72.0 \text{GNm}^{-2}) \). Both the experimentally determined and theoretically derived stresses were used to calibrate the SPATE data.

4.3 Results

The thermoelastic constant for human enamel was calculated theoretically using the following formula:
<table>
<thead>
<tr>
<th>Restoration</th>
<th>Specimen</th>
<th>Sum of Principal Stresses</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Buccal Surface</td>
<td>Lingual Surface</td>
</tr>
<tr>
<td>Unprepared tooth</td>
<td>A</td>
<td>-1.2 - 2.8</td>
<td>-1.2 - 3.6</td>
<td></td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>-1.2 - 1.2</td>
<td>-1.2 - 2.8</td>
<td></td>
</tr>
<tr>
<td>Prepared tooth</td>
<td>A</td>
<td>-2.8 - 5.2</td>
<td>-3.6 - 5.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>-3.6 - 5.2</td>
<td>-3.6 - 5.2</td>
<td></td>
</tr>
<tr>
<td>Amalgam restn.</td>
<td>A</td>
<td>-2.8 - 5.2</td>
<td>-3.6 - 5.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>-3.6 - 5.2</td>
<td>-3.6 - 5.2</td>
<td></td>
</tr>
<tr>
<td>Composite restn.</td>
<td>A</td>
<td>-1.2 - 1.2</td>
<td>-1.2 - 2.8</td>
<td></td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>-1.2 - 1.2</td>
<td>-1.2 - 1.2</td>
<td></td>
</tr>
</tbody>
</table>

Table 10. The range of the sum of the principal stresses measured using the thermoelastic method on the surface of intact and restored teeth subjected to an occlusal load of 200N.

Figure 23. Range of the principal stresses observed on the surface of teeth subjected to a range of operative procedures by measurement of thermoelastic emission.
\[ K_m = \frac{\alpha_T}{p C_p} \]

where

- \( K_m \) = Thermoelastic constant
- \( \alpha_T \) = Coefficient of thermal expansion
  
  \((11.4 \times 10^{-6} \, \text{K}^{-1} \text{ for human enamel [Braden, 1976]})\)
- \( p \) = density
  
  \((2980 \, \text{kg/m}^3 \text{ [Weidmann et al., 1967]})\)
- \( C_p \) = Specific heat at constant pressure
  
  \((170 \, \text{J/kg K [Brown et al., 1970]})\)

The resulting value for \( K_m \) of \(2.25 \times 10^{-12} \, \text{m}^2/\text{N} \) was used to calibrate the SPATE plots in terms of the sum of the principal stresses. Table 10. and Figure 23. illustrate the range of the principal stresses observed on the buccal and lingual surfaces of the two specimens and variations resulting from different operative procedures. Values given are the sum of the principal stresses in MNm\(^{-2}\). A calibrated SPATE plot of the sum of the principal stresses on the buccal surface of an unprepared tooth (specimen A) is shown in Figure 24.(a). The distribution of stress along an occluso-cervical profile of this plot is represented graphically along a line of high stress in Figure 24.(b). Further plots illustrate the distribution of stresses over the buccal surface of the same specimen (A) following preparation (Figure 25.(a)) and restoration with a bonded composite restoration (Figure 27.(a)). Occluso-cervical profiles were also produced through regions of high stress and are represented graphically in figures 25.(b) and 27.(b). A plot of the lingual surface of specimen (A) following cavity preparation is shown in figure 26.(a).

Different specimens appeared to have a similar pattern of stress distribution when plots were compared for the same restorative procedures. Unprepared specimens exhibited a comparable magnitude and distribution of stresses to those seen when teeth were restored with enamel etched and dentine bonded composite material. There were also similarities between the plots for prepared specimens and those restored with amalgam restorations.

Values for the principal stresses calculated from strain gauge measurements made in
Figure 24. (a) SPATE plot of the principal stresses on the buccal surface of an unprepared tooth subjected to a dynamic load of 200N

(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of an unprepared tooth
Figure 25. (a) SPATE plot of the principal stresses on the buccal surface of a prepared tooth subjected to a dynamic load of 200N

(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of a prepared tooth
Figure 26. (a) SPATE plot of the principal stresses on the lingual surface of a prepared tooth subjected to a dynamic load of 200N
(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of a prepared tooth
Figure 27.(a) SPATE plot of the principal stresses on the buccal surface of a tooth restored with a bonded composite material and subjected to a dynamic load of 200N.

(b) Principal stresses plotted against distance for an occluso-cervical profile of the SPATE plot of a tooth restored with a bonded composite material.
the pilot study were 0.56MNm\(^{-2}\) for \(\sigma_1\) and -0.12MNm\(^{-2}\) for \(\sigma_2\). The sum of the principal stresses on the surface of the specimen in close proximity to the strain gauge was therefore 0.44MNm\(^{-2}\). This was in close agreement with a value of 0.40MNm\(^{-2}\) recorded from an area of a theoretically calibrated SPATE plot of the same specimen in close proximity to the strain gauge.

### 4.4 Discussion of the method and results

The thermoelastic constant \((K_m)\) for human enamel was calculated using values for the coefficient of thermal expansion \((\alpha_T)\), density \((p)\) and specific heat \((C_p)\) at constant pressure reported in the literature (Braden, 1976). These properties have not been widely studied and only one reference was found for each constant. The specific heat of enamel reported in the literature (Brown et al., 1970) was determined at atmospheric pressure and the material allowed to expand as it was heated. The value quoted is therefore at constant pressure \(C_p\), which is a zero stress condition. Harwood and Cummings (1986) suggested that volume changes occurring in a material subjected to loading may also influence the value of the specific heat. However, they showed that the effective specific heat \((C_e)\) which took this into account resulted in a value for the specific heat for mild steel which was only fractionally lower than \(C_p\). It was not therefore considered necessary to make compensations in the calculation of \(K_m\) for human enamel. The value of \(K_m\) for enamel, \(2.25 \times 10^{-12} \text{m}^2\text{N}^{-1}\), compared favourably with an experimentally and theoretically derived value for steel of \(3.5 \times 10^{-12} \text{m}^2\text{N}^{-1}\) (Harwood & Cummings, 1986).

Figure 23 and Table 10 illustrate the range of the sum of the principal stresses for the different specimens and restorative procedures and suggest a reasonable correlation between the different specimens subjected to the same restorative procedure which was supported by the pattern of stress distributions in the different plots. It was interesting to note that the principal stresses increased greatly on loading prepared but unrestored specimens. The magnitude and distribution of stresses did not alter significantly on placement of an amalgam restoration which suggests that the restoration only plays a passive role. The magnitude of the sum of the principal stresses was much lower and the stress distribution quite different on replacement of the amalgam restoration with an
etched and bonded composite restoration. Variations in the distribution of stresses was evident in plots of the buccal surface of specimen A, intact, prepared and restored with the composite material (Figures 24, 25, 27). It appears therefore, that cavity preparation and placement of bonded composite restorations may alter the stress distribution in extracted teeth subjected to axial loading in-vitro. However, it is important to consider that the thermoelastic technique only measures stresses resulting from direct application of a dynamic load and is not able to resolve residual stresses such as those that may result from polymerisation of an enamel etched and dentine bonded composite restoration (Meredith & Setchell, 1987; Causton, Miller & Sefton, 1985).

The distribution of stresses on the graphic plots indicated that the regions of highest stress on the prepared specimens were in the mid-buccal portion of the tooth level with or just below the floor of the cavity. Plots made along an occluso-cervical section (Figures 24, 25, 27 (b)) through this region enabled the distribution to be seen more clearly. The levels of the maximum stress were considered to be high at 5.2MNm$^{-2}$ but the applied load, 200N, is probably higher than would be seen in clinically in function. The wide variations in the direction and magnitude of occlusal loading in-vivo should also be considered.

Stress distributions on the plots were different for buccal and lingual surfaces c.f. Figure 25(a) and 26(a) however, the overall magnitudes of stress did not appear to differ greatly.

There was good correlation between the stresses calculated from strain gauges placed in a region of uniform stress on a specimen surface with those based on the theoretical calculation of $K_m$ for enamel. This calibration technique followed the method recommended by Harwood and Cummings (1986) and the results indicated that the thermo-elastic technique may be applied to dental hard tissues with accurate results. Potential difficulties may arise however, in carrying out measurements on a specimen that is composed of two different materials (i.e. enamel and gold at a crown margin) which have different values of $K_m$ as this will make calibration difficult.

This investigation has shown that restorative procedures can influence the distribution of stresses in teeth subjected to loads in-vitro. Areas of high stress have been identified
but a more detailed measurement technique such as strain gauges should be used to determine the magnitude and direction of the principal strains on teeth subjected to a range of loading conditions and determine changes resulting from placement of restorations.
CHAPTER 5

IN-VITRO DISCRETE FIELD STRESS ANALYSIS OF INTACT AND RESTORED TEETH AND MODEL SYSTEMS USING STRAIN GAUGES.

5.1 Aims and objectives

5.2 Method

5.2.1 Specimen preparation - natural teeth

5.2.2 Specimen fabrication - composite model system

5.2.3 Cavity preparation - natural teeth

5.2.4 Cavity restoration

5.2.5 Instrumentation

5.2.5.1 Load application and control

5.2.5.2 Loading rig - design and fabrication

5.2.5.3 Loading rig - compliance

5.2.5.4 Strain measurement

5.2.5.5 Strain gauge instrumentation

5.2.5.6 Data acquisition

5.2.6 Test Procedure

5.3 Results

5.3.1 Calibration

5.3.2 Restorative procedures

5.3.3 Loading procedures

5.4 Discussion

5.4.1 Discussion of the method

5.4.2 Discussion of the results

5.4.2.1 Restorative procedures

5.4.2.2 Loading procedures
5.1 Aims and objectives

A number of retrospective clinical studies (Hansen, 1988; Bell, Smith & DePont, 1982) have established that intact and restored teeth may fracture clinically under certain loading conditions. Unfortunately, it is very difficult to measure occlusal loads and resultant stresses in-vivo and predict the magnitude and direction of force necessary to cause tooth fracture. The limitations of measuring the fracture resistance of teeth in-vitro and the clinical significance of the results have already been discussed (see 1.7.5).

Results from a non-contacting in-vitro thermoelastic stress analysis technique have identified regions of high stress on intact and restored teeth which may be sites of potential failure. These findings warrant further investigation and placement of strain gauges on those areas exhibiting high levels of stress will enable the magnitude and direction of the principal strains to be deduced.

Evidence from the thermoelastic analysis suggested that preparation and restoration of teeth may alter the stress distribution on subsequent load application. However, this technique did not measure stresses produced in the remaining tooth structure during preparation or placement of a restoration. It has been shown that polymerisation of an acid etched and dentine bonded composite restoration produces measurable changes in strain in the remaining tooth structure (Meredith & Setchell, 1987) and further work is necessary to establish the relationship between stresses on a tooth surface and the type of restoration, the material used and the influence of subsequent loading.

Small size, irregular geometry and complex physical properties have made study of the mechanical behaviour of human teeth very difficult. If the effects of some of these variables could be reduced, by using a model system to simplify geometry and material properties for example, it may be possible to identify the mechanical interactions between a tooth and restoration more clearly. Potential model systems were studied alongside natural teeth in this investigation. The first type of model was fabricated from a relatively homogeneous isotropic composite material(12) and had an angular geometrical shape (Figure 28.) representing a molar tooth prepared with a mesio-distal cavity. The second model fabricated from the same material had a more complex geometry and was molariform in shape again containing a cavity.
Figure 28. Dimensions of geometric composite models
It was considered that loading parameters including amplitude, rate and position also warranted investigation and the application of relatively low loads resulting in elastic deformation without causing specimen damage could enable specimens to be tested a number of times using different restorative procedures.

In almost all the published in-vitro work investigating the fracture resistance of restored teeth specimens have been subjected to axial occlusal loads representing intercuspal contacts. However eccentric, working and non-working side contacts may occur clinically in lateral excursions of the mandible and these should also be investigated.

5.2 Method

5.2.1 Specimen preparation - natural teeth

Criteria for specimen type, selection and storage was the same as that used in the previous investigation (see 4.2.1). Prior to mounting specimens in brass rings strain gauges were attached to the buccal and lingual surfaces.

The strain gauges selected were three element 45° rosettes having a grid length of 1mm and width of 0.7mm(41). Thermal characteristics of the gauges were matched with those of the specimens by selecting strain gauges having a value for the coefficient of thermal expansion \((11 \times 10^{-6})\) close to that of enamel. Integral leads facilitated easier gauge mounting. Specimens were prepared by lightly etching areas approximately 5mm in diameter on the buccal and lingual surfaces in the middle of the crown of each tooth for fifteen seconds with a phosphoric acid gel(36). Following washing and drying a strain gauge was positioned on the buccal surface of each specimen with the aid of a stereo-microscope at 12x magnification(20) and bonded using a cyanoacrylate based adhesive(21). Tactile pressure was applied through a polyacetate film for a period of two minutes to ensure good adaptation and minimise the adhesive film thickness. Each gauge was placed at the mid-point mesio-distally, level with the floor of the cavity on each specimen and orientated so that the central element (3) lay perpendicular to the occlusal table and parallel to the long axis of the tooth. The other two elements on the rosette were angulated at 45° mesial (1) and distal to this (2) (Figure 29). To aid gauge alignment and
Figure 29. Rosette strain gauge mounted on the buccal surface of a lower first molar and indicating the level of the cavity floor.
establish the relationship between the buccal and lingual gauge specimens were mounted in a rig that used a pair of mirrors at fixed angles to enable the occlusal, buccal and lingual surfaces of the tooth to be viewed simultaneously. The lingual gauge was placed in the same mesio-distal and occluso-cervical axis as the buccal gauge and was orientated at the same angle (Figure 30.). Bond quality was checked visually and tested with a scalpel blade. Following placement, gauges were protected against moisture contamination by application of a thin coating of an air drying acrylic resin (22).

Teeth were mounted in brass rings (30) and embedded using a technique similar to that described previously (see 4.2.1). The lead wires from the strain gauges were curled part of the way into the mounting ring and became rigidly attached following embedding. This provided an element of strain relief for the wires. Mounted specimens were numbered and stored under water at 4°C until ready for testing.

5.2.2 Specimen fabrication - composite model system

Specimens were fabricated from a chemically curing composite material (12), the physical properties of which had been measured in an earlier part of the study (see 3.4.2.1). One type of model was angular and of average dimensions for a lower permanent molar tooth (10mm mesio-distally x 9.75mm bucco-lingually x 7.5mm occluso-cervically; Wheeler, 1971 [Figure 28.]) containing a mesio-distal cavity. The other model had a more complex, molariform shape and again contained an MOD type cavity. A machined aluminium die and melamine lower left first molar tooth (Columbia Dentoform; USA [42]) were used as master models to fabricate silicone moulds for the replication of the composite specimens.

The composite material (12) was mixed using a technique similar to that described previously (see 3.2.2). Each specimen was allowed to polymerise at an ambient temperature of 21°C ± 1°C for a period of ten minutes and then removed from the mould. The internal surface of the mould partially inhibited polymerisation of the composite resulting in an incompletely cured surface layer approximately 200μm thick. This was removed by rubbing with a cotton pledget soaked in a 10% solution of methyl alcohol. Ten specimens of each type were fabricated, numbered and stored at an ambient temperature and humidity for a period of not less than fourteen days prior to testing to
Figure 30. Relationship between buccal and lingual strain gauge and orientation relative to the occlusal plane.
allow for stress relaxation in the polymer.

Strain gauges were bonded to the buccal and lingual cusps of each specimen using a technique similar to that described earlier (5.2.1) however, the specimen surface was not etched or degreased but cleaned by application of a 10% solution of methyl alcohol. Testing was not performed on specimens until a minimum of thirty minutes following gauge placement. Mounting and embedding specimens was also carried out using a method similar to that described for natural teeth. Silicone jigs were used to aid specimen positioning in the brass rings resulting in a standardised orientation.

5.2.3 Cavity preparation - natural teeth

As part of the testing procedure the natural teeth were prepared with occlusal and subsequently mesio-occluso-distal (MOD) cavities. This was performed with a standardised technique using a pear shaped tungsten carbide bur (32) in a high speed dental handpiece under constant lubrication. Cavity dimensions were kept as constant as possible between specimens. MOD cavities were slot preparations and there was no narrowing of the isthmus or presence of mesial or distal boxes. Bucco-lingual cavity width was approximately one third of the width of the specimen and a steel ball bearing 4.5mm in diameter formed a three point contact with the remaining tooth structure in the occlusal fossa (Figure 31.). Testing was performed immediately following preparation and cavities were kept moist throughout the loading procedure with a dampened cotton wick.

5.2.4 Cavity restoration

Cavities in both the natural teeth and the composite models were restored in sequence using a number of materials and techniques;

a) Occlusal cavities in natural teeth were restored with an amalgam restoration and subsequently an enamel etched and dentine bonded directly placed composite restoration.

b) MOD type cavities in the two types of composite model system were restored with an amalgam restoration and subsequently an enamel etched and dentine bonded directly placed composite restoration.

c) MOD cavities in natural teeth were restored with an amalgam restoration, an
Figure 31. In-vitro loading positions on natural teeth and composite specimens.
enamel etched and bonded directly placed composite restoration, a directly placed composite restoration with a light polymerising glass-ionomer lining and an indirectly fabricated enamel etched and dentine bonded composite restoration.

Amalgam and etched and bonded composite restorations were placed using techniques similar to those previously described (see 4.2.1.2). Cavity margins in the composite models however, were not acid etched and a priming agent was not used prior to application of the light curing dentine adhesive. Care was taken during specimen restoration that the matrix band did not impinge on the strain gauges. No significant changes in cuspal strain were observed during band placement. This is in contradiction to a study by Grimaldi and Hood (1973) who reported cuspal flexure of up to 80μm on tightening a matrix band but may be explained by the type of matrix used (33) which has a clutch in the handpiece to prevent overtightening.

Restoration of MOD cavities with a glass ionomer lining and composite material was performed in two stages: A light activated glass ionomer material (3) was mixed in accordance with the manufacturer's instructions and placed on the floor of each cavity to an approximate depth of 1mm. The material was polymerised by exposure to a visible light source (14) for twenty seconds. The enamel cavity margins were then etched and any remaining exposed dentine was primed with the priming agent. A light curing dentine adhesive was applied to the primed dentine surface and glass ionomer lining and exposed to a visible light source for twenty seconds. In the second stage the remaining part of the cavity was restored with a composite material (11) using the technique described previously (see 4.2.1.2).

MOD type cavities in natural teeth were also restored with a composite restoration placed using an 'indirect' technique in which the material was polymerised out of the cavity and then cemented in place using a luting cement (Experimental material LC; 3M Co. Loughborough, England [33]). This was carried out in three stages;

i) A thin silicone oil acting as a release agent was sprayed over the cavity surface. The light curing composite material used previously (11) was placed in each cavity in a single increment and contoured to restore the original anatomy of the tooth before
Figure 32. Timing diagram illustrating rate and magnitude of load application to specimens
polymerising with a visible light source\textsuperscript{(14)} for thirty seconds. The composite restoration was then carefully removed from the cavity.

ii) Each of the six surfaces of the partially polymerised composite restoration was exposed to a visible light source for a period of forty seconds to produce a high level of conversion in the material.

iii) The internal surfaces of the composite restoration and the cavity walls and floor were thoroughly washed and dried to remove traces of silicone oil. The enamel margins were then etched and a bonding agent applied to the dentine in the manner described previously. An experimental composite luting cement composed of two pastes\textsuperscript{(143)} was mixed in accordance with the manufacturers instructions and a thin coat applied to both the margins of the cavity and fitting surface of the inlay. The composite inlay was then seated under pressure to extrude excess cement and ensure a minimal film thickness. Excess material was wiped from the margins and polymerisation of the cement initiated by exposure to a visible light source for twenty seconds. An abrasive alumina stone mounted in a high speed handpiece was used to smooth the restoration margins.

Specimens restored with composite materials were stored for a minimum of two hours at an ambient temperature and humidity prior to testing.

5.2.5 Instrumentation

5.2.5.1 Load application and control

Loads designed to simulate those seen clinically in function were applied to the natural teeth and composite models. A uni-axial servo-hydraulic testing machine\textsuperscript{(38)} was used to generate loads which were transmitted to the specimen surface via a loading rig. The load cycle was programmed on a micro-computer connected to the testing rig via a serial interface and specimens were subjected to loads of 10, 20, 50, 100 and 200N at three loading rates 0.01, 0.05 and 0.10KNsec\textsuperscript{-1}. A loading cycle comprised an initial 'rest' period of 3.5 seconds at no load, a loading ramp, a static period of 3.5 seconds at maximum load and lastly an unloading ramp at the predetermined rate (Figure 32.).

A range of loading positions represented an inter-cuspal contact and occlusal contacts in lateral excursions of the mandible (see Figure 31.). A loading rig was designed to
Figure 33. 3rd angle projection of specimen loading rig
enable adjustments in specimen position to accommodate the range of contacts and variations in specimen geometry.

The 20KN load cell and 2KN sub-cell fitted to the servo-hydraulic testing machine were calibrated electronically but manual adjustment of the zero offset was possible. It was considered that calibration accuracy and linearity of the load cell and A-D converter should be checked periodically by means of a series of deadweight loads\(^{16}\). Accuracy and linearity were better than \(\pm 0.5\%\) of the measured load. A slight disadvantage in the use of the servo-hydraulic machine was the need for a preload of 2-3N to be applied to the specimen prior to testing to provide a feedback signal for the control system.

5.2.5.2 Loading rig - design and fabrication

The objectives in designing the loading rig were to enable specimens to be translated and rotated mesio-distally and bucco-lingually. The loading rig was based on a design by Kempson (1970) used to indent cartilage. Figure 32 illustrates a third angle view of the rig taken from working drawings and Figure 33. a photograph of the specimen mounting assembly. The frame (with reference to Figure 32.) \(^{2}\) was fabricated from aluminium alloy channel which supported a swinging base\(^{10}\) on which was mounted the x-y slides and specimen block\(^{11-17}\). The swinging base was pivoted on brass bearings mounted in the side frames enabling the angle of load application to be adjusted and the base locked into position. The mounting block\(^{17}\) supported the brass ring in which the specimen was embedded and this was bolted to a variable x-y slide\(^{12-15}\) which enabled mesio-distal and bucco-lingual specimen to be altered using two micrometer screws with a resolution of 10\(\mu\)m. Engravings on the mounting block enabled accurate repositioning of specimens.

The main load bearing assembly\(^{19}\) was fabricated from an aluminium alloy shaft in a brass bush and mounted on a horizontal cross-piece. An LVDT with linear bearings\(^{17}\) also fitted to the cross-piece was connected in parallel to the bearing by means of a small aluminium bar to measure bearing displacement. The load was transmitted from the load cell mounted on the cross-head of the testing machine to the specimen via a ball bearing assembly\(^{21}\) which compensated for any eccentricity between the testing machine and the rig.
Figure 34. Photograph of specimen mounting assembly and method of load application (Load position 1)
5.2.5.3 Loading rig - compliance

Stiffness and rigidity were important design considerations in development of the loading rig. The major measurement parameter was strain monitored using gauges mounted on the buccal and lingual cusps. Although deflection of the rig was considered important only if it resulted in a change in load direction or point of application it was felt that rig compliance should be assessed and this was carried out in three stages;

i) Axially directed and angulated (30°) loads were applied via the load bearing assembly to a mild steel block which replaced the swing base\textsuperscript{[10]}. 

ii) Axially directed and angulated (30°) loads were applied via the bearing to a steel block which fitted in the specimen mounting block mounted on the swinging base.

iii) Axially directed and angulated (30°) loads were applied to a geometrical composite specimen embedded in a brass ring in the mounting block.

5.2.5.4 Strain measurement

Strains were recorded from the three elements of the buccal and lingual strain gauges on each specimen during restorative procedures and subsequent loading. During restorative procedures embedded specimens were mounted in a machined perspex block to ease handling. A control specimen, to which strain gauges had been attached, was also mounted in the block. Each element of the buccal and lingual strain gauges mounted on the test specimens was connected as the active arm of a Wheatstone bridge circuit and gauges on the control specimen comprised the dummy arms of the circuit and compensated for fluctuations in temperature. It was considered that temperature changes may pose a problem when a specimen was exposed to a visible curing light as part of a restorative procedure. Therefore two curing lights were used simultaneously, one illuminating the test and the other the control specimen.

Although specimen loading procedures were performed at an ambient temperature and humidity it was thought that air currents may influence gauge readings. A perspex enclosure was therefore designed and constructed covering the entire loading rig. The temperature of the air inside the enclosure was regulated electronically and heated by a
Figure 35. Schematic for specimen loading and strain measurement instrumentation
500 watt halogen light source circulated by a variable speed fan. In practice it was found that temperature regulation was not necessary as the variation in ambient temperature during testing procedures was less than one degree.

Changes in buccal and lingual strain were measured on specimens during all loading procedures. The strain gauges mounted on the specimens formed the active arms of the measurement bridges and dummy gauges bonded to small pieces enamel or composite were positioned in close proximity to the specimens to compensate for variations in temperature. The dummy gauges were mechanically decoupled from the test specimens and mounting rig by using adhesive foam pads to minimise stresses during loading.

Displacement of the load bearing was monitored during loading procedures by means of the LVDT mounted on the rig to ensure that the loading point did not slip on the tooth surface resulting in a change in direction of the load or point of application; a phenomenon which may have been difficult to observe from the strain gauge readings alone.

5.2.5.5 Strain gauge instrumentation

Each element of the buccal and lingual rosette gauges (six in total) were connected as part of a Wheatstone bridge to a strain gauge amplifier which had been custom built for the purpose (see Appendix III). Each gauge had a resistance of 120Ω and was excited by a D.C. voltage of 0.5V. This provided sufficient sensitivity but minimised heat dissipation from the gauge surface.

The gauge excitation voltage and amplifier output for each channel was monitored prior to each test run and a shunt resistor was switched in parallel to the active element of each bridge simulating a mechanical strain of 1000με (Figure 35.). Amplifier gain (gauge factor) was kept constant throughout the experiments but the amplifiers were zeroed where necessary to prevent over-rangeing.

5.2.5.6 Data acquisition

Load, strain gauge, displacement and timing data was monitored and logged using a sixteen channel, twelve-bit analogue-to-digital converter(24) connected to a microcomputer(25). The following channel allocation was adopted;
Figure 36. Sequence of load application, restorative procedures and strain measurement
1) Element 1 - Buccal strain gauge
2) Element 2 - Buccal strain gauge
3) Element 3 - Buccal strain gauge
4) Element 3 - Lingual strain gauge
5) Element 2 - Lingual strain gauge
6) Element 1 - Lingual strain gauge
7) Load cell output, testing machine
8) LVDT output
9) Timing pulse

Pilot studies were performed to enable the sampling rate and hence the amount of data collected to be optimised for the restorative and loading procedures. It was found that the minimum sampling rate of 9Hz attainable using the software package provided with the converter\textsuperscript{(24)} was too fast and produced too much data making storage inefficient and analysis time-consuming. The commercial software was therefore rewritten to enable sampling rates from 0.0001 to 20,000 Hz to be obtained. As a result sampling rates of 3Hz and 5Hz were chosen for the restorative and loading procedures providing the optimum combination to record changing trends and remain economical on data storage. Results for each restorative or loading procedure were stored as an ASCII file on disc which contained nine columns of data in addition to the sample number.

The analogue-to-digital converter converted the analogue voltages from the transducer amplifiers into 4byte digital values using a 12-bit successive approximation method; data for each transducer was thus stored as a four digit number in the range 0-4096. The input range of the A-D converter was ±5V and the gain of the transducer amplifiers was adjusted so that measurements did not exceed this.

5.2.6 Test Procedure

Pilot studies were carried out to establish optimum loading parameters and to ensure that these did not cause specimens to fracture or undergo irreversible physical changes. The restorative procedures were performed in accordance with the methods described previously (see 5.2.3 & 5.2.4) and in the sequence illustrated in Figure 36. Care was
Figure 37. Load-displacement plots for different specimen types subjected to axial and angular loads.
taken that operative procedures were performed in a standardised manner and removal of restorations was carried out with minimal trauma and removal of remaining tooth structure. All the stages were performed on the intact natural teeth but only stages 11-15 and 20-22 were carried out on the composite models systems.

Specimens were loaded at five positions on the surfaces of the natural teeth and six on the composite models (Figure 31.). The position, direction, and type of contact for loads applied to the natural teeth and molariform composite specimens were:

[1] An axially directed contact between the occlusal fossa and a steel bearing 4.5mm diameter forming a tripod contact on natural teeth and a two-point contact on the geometric composite model, simulating a functional intercuspal contact.

[2] A 30° angulated contact on the buccal incline of natural teeth and composite specimens formed by a 1mm diameter steel pin and simulating a working side contact in lateral mandibular excursions.

[3] A 30° angulated contact of 1mm diameter on the buccal incline of the lingual cusp on natural teeth and composite specimens simulating a working side interference in lateral excursions.

[4] A 30° angulated contact 1mm in diameter on the lingual incline of the buccal cusp on natural teeth and composite specimens simulating a non-working side interference in lateral excursions.

[5] A 30° angulated contact, 1mm in diameter on the lingual incline of the lingual cusp on natural teeth and composite specimens simulating a non-working side interference in lateral excursions.

Loading positions on each specimen had been marked with an articulating foil\(^{(39)}\) and this enabled specimen repositioning to be carried out to an estimated accuracy of within ±200\(\mu\)m.

Buccal and lingual cuspal strains were recorded during the loading and restorative procedures as outlined in Figure 36. Sampling periods of 4.5 and 10 minutes at rates of 5Hz and 3Hz were used to log data during the loading and restorative procedures.
Figure 38. Frequency response of rosette type strain gauge mounted on a composite specimen and subjected to 100N load at different rates.
5.3 Results

5.3.1 Calibration

The results of the compliance tests illustrated in Figure 37. indicate that the steel specimens exhibited near linear, elastic behaviour when subjected to axial loading and unloading. An increase in seating marked by a shift in displacement was evident on the composite specimen during the loading cycles. There was some hysteresis during angled unloading of both steel and composite specimens. This was probably due to the edge of the flat of the loading tip binding against the specimen because of lateral thrust from the main load bearing. It was found that slightly rounding the margins of the tip reduced this problem.

The frequency response of the strain gauges and amplifiers were measured to ensure that they were capable of responding to the rapid changes in stress in a specimen during loading. Figure 38. illustrates the response of each element of a rosette gauge mounted on the buccal surface of a composite specimen which was subjected to an intercuspal load of 100N at rates varying from 0.01-0.16KNsec⁻¹. The error in attaining the peak strain at any loading rate was calculated as less than ±5με and this was considered to be satisfactory.

It was also considered important that the drift in output of the strain gauge amplifiers should be assessed in relation to temperature and time. Figure 39. illustrates strains measured by buccal and lingual rosette gauges mounted on a natural tooth over periods of twenty minutes and twelve hours. It can be seen that in both cases the temperature measured in close proximity to the gauges inside the environmental enclosure varied by less than ±0.5°C. Drift of the measured strain which was generally less than ±10με over periods of twenty minutes was considered to be satisfactory but became more significant over longer periods and the relatively small changes in ambient temperature suggested that this may be due to thermal instability in the amplifier rather than the strain gauges.

Calibrated values of strain were calculated for each element of the buccal and lingual rosettes from the raw digital data according to the formula;

\[ \epsilon_s = \frac{1000}{(D_{1000} - D_0) \times (D_M - D_0)} \]  

5.1
Figure 39. Apparent strain in relation to time and temperature over periods of twenty minutes and twelve hours.
where:

\[ \epsilon = \text{Absolute strain in microstrain units (\(\mu\epsilon\))} \]

\[ D_{1000} = \text{Digital strain reading during 1000\(\mu\epsilon\) calibration pulse} \]

\[ D_0 = \text{Digital value at zero strain} \]

\[ D_M = \text{Digital value at measured strain} \]

The three individual strain readings for each element on the buccal and lingual rosettes were then used to calculate the following parameters;

a) the maximum and minimum principal strains for buccal and lingual cusps
   
   \((B_{e1}, B_{e2}, L_{e1}, \text{and } L_{e2})\),

b) the maximum shear strains \((B_r, \text{and } L_r)\),

c) the angle of the principal and shear strains with the respect to the angle of
   element [1] on the buccal and lingual rosettes \((B_{11, 1}, B_{12, 1}, B_{1a}, L_{11, 1}, L_{12, 1}, \text{and } L_{1a})\).

Data was calculated using the computer program listed and described in appendix IV. The relationships between individual element readings and overall value of principal strains can be seen in Figure 40, which illustrates the loading cycle and resultant strains for a natural tooth prepared with an MOD cavity and subjected to axial inter-cuspal loading (load position [1]). An example of the magnitude and directions of the principal and shear strains on the surface of an intact tooth subjected to an intercuspal load of 100N is shown graphically in Figure 41.

Principal strains were also used to derive values for the principal stresses, \(B_{\sigma 1}, B_{\sigma 2}, L_{\sigma 1}\) and \(L_{\sigma 2}\):

\[ B_{\sigma 1} = \frac{E}{1-\nu^2} (B\epsilon_1 + \nu B\epsilon_2) \]

\[ B_{\sigma 2} = \frac{E}{1-\nu^2} (B\epsilon_2 + \nu B\epsilon_1) \]

using a range of values for Young’s modulus of Enamel (E) taken from the literature (24.8, 40.6 and 81.0 GNm\(^{-2}\)) and a range of values substituted for Poisson's ratio (\(\nu\)) of 0.25-0.30.
Figure 40. Derivation of principal and shear strains in a specimen prepared with an MOD cavity and subjected to a range of intercuspal loads at different rates.
Figure 41. Graphical representation of the magnitude and direction of the principal and shear strains on the buccal and lingual surfaces of an intact tooth subjected to an intercuspal load of 100 Newtons.
Figure 42. Changes in strain during placement and polymerisation of a bonded composite restoration in a molar tooth
Table 11. Mean changes in principal and shear strains during placement of occlusal and MOD type composite restorations in natural teeth
5.3.2 Restorative procedures

Changes in cuspal strain were measured during the restorative procedures outlined in Figure 36. It was found that changes in strain during placement of occlusal amalgams in natural teeth (procedure [4]) and MOD amalgams in natural teeth and the composite model systems [13] were relatively small and generally fell within a range of ±10με which was considered to be attributable to experimental error.

Measurement of strains during placement and polymerisation of acid etched and dentine bonded composite restorations in natural teeth and the composite models were divided into a number of steps. Figure 42. illustrates the changes in individual element (a) and principal (b) strains during placement of an MOD composite restoration in a natural tooth [20]. Mean changes in principal and shear strains for each stage during placement of occlusal [7] and MOD type composite [20] restorations in natural teeth and the molariform composite models are given in Table 11. Total strains are given for these procedures and also the glass ionomer and composite [17] and indirect composite [24] restorations in Table 12. Data in Table 11 is presented graphically in Figure 43.(a),(b) and (c) to illustrate trends more clearly.

A multiple comparison of means was performed using a generalised linear modelling procedure (PROC GLM) to compare differences in the total principal and shear strains for the different restorative procedures. The results indicated that there were no interactions between the specimen type and restorative procedure. The test revealed significant differences at the 5% level between the different restorative procedures for the buccal and lingual shear strains. A multiple comparison of means performed using Bonferroni’s t test (Miller, 1981; PROC GLM(27)) revealed significant differences (Pr<0.05) in the maximum buccal and lingual shear strains between natural teeth restored with indirect (MOD) composite restorations [24] and those restored with composite [20] and glass ionomer and composite [17] restorations.

5.3.3 Loading procedures

The resultant strains, experimental loading variables, restoration variables and specimen types were stored as a series of tables in a normalised database (third normal
### Total mean strains during restoration placement

<table>
<thead>
<tr>
<th>Restoration</th>
<th>Buccal Max. strain</th>
<th>Conf. Intvl. 95% +/−</th>
<th>Buccal Min. strain</th>
<th>Conf. Intvl. 95% +/−</th>
<th>Lingual Max. strain</th>
<th>Conf. Intvl. 95% +/−</th>
<th>Lingual Min. strain</th>
<th>Conf. Intvl. 95% +/−</th>
</tr>
</thead>
<tbody>
<tr>
<td>Directly placed MOD composite</td>
<td>-51</td>
<td>156</td>
<td>82</td>
<td>108</td>
<td>-244</td>
<td>60</td>
<td>61</td>
<td>47</td>
</tr>
<tr>
<td>Directly placed glass ionomer lined composite</td>
<td>-40</td>
<td>46</td>
<td>67</td>
<td>87</td>
<td>-180</td>
<td>105</td>
<td>92</td>
<td>62</td>
</tr>
<tr>
<td>Indirectly placed MOD composite</td>
<td>-15</td>
<td>14</td>
<td>22</td>
<td>15</td>
<td>-63</td>
<td>82</td>
<td>52</td>
<td>84</td>
</tr>
</tbody>
</table>

Table 12. Total strains produced during placement of direct composite, indirect composite and glass ionomer lined composite restorations in natural teeth
Figure 43.(a) Mean cumulative principal and shear strains during placement of MOD bonded composite restorations in composite models
Occlusal composite restorations in natural teeth

![Graph showing strain vs. procedure for occlusal composite restorations.]

MOD composite restorations in natural teeth

![Graph showing strain vs. procedure for MOD composite restorations.]

(b) Mean cumulative principal and shear strains during placement of occlusal bonded composite restorations in natural teeth

c) Mean cumulative principal and shear strains during placement of MOD bonded composite restorations in natural teeth
Figure 44. Mean strains for teeth prepared with MOD cavities and subjected to intercuspal loads.
form; Paradox v.2.1; Borland International (UK)[44]) An index field linked these tables enabling rapid searching, sorting and filtering to be performed.

Loading parameters included amplitude, rate and position. A test for normality performed on the loading data as part of a univariate procedure (Shapiro-Wilkes; PROC UNIVARIATE)\(^{27}\) suggested that the data did not exhibit a normal distribution. In view of the small sizes of specimen groups (n=7) non-parametric statistics were performed except where comparison of a number of different parameters was carried out when a generalised linear modelling procedure was used, because in some cases (due to gauge damage) groups were of different sizes.

A non-parametric analysis of variance (Friedman test)\(^{27}\) revealed that there were no significant differences at the 5% level in the peak strains reached at loading rates of 0.01, 0.05 and 0.1 KNsec\(^{-1}\) in natural teeth prepared with MOD type cavities and loaded intercuspally (position [1]) or at an angle on the buccal cusp [2,4]. No differences were observed for restored teeth and subsequent analyses were performed on data taken from specimens loaded at 0.1 KNsec\(^{-1}\) which was considered to be representative of the rate of loading in-vivo.

There appeared to be a near linear relationship between the loading amplitude and principal and shear strains. This can be seen in Figure 44. which illustrates the mean strains for natural teeth prepared with MOD type cavities and loaded intercuspally [1]. Although confidence intervals for the different loading amplitudes at the 5% level in Figure 44. appear relatively large, a generalised linear modelling technique (PROC GLM)\(^{27}\) used to compare amplitudes and specimen types (i.e. if the tooth was intact, prepared or restored) indicated a significant difference in strains at the different amplitudes. There was no evidence of interactions between the specimen type and load. This test also revealed significant differences at the 5% level between the different specimen types for the maximum buccal and lingual shear strains.

A multiple comparison of means using Bonferroni’s t test ([Miller, 1981]; PROC GLM)\(^{27}\) revealed that there were significant differences (Pr < 0.05) between the maximum buccal shear strains in intact natural teeth and those containing occlusal cavities and between intact teeth and those containing MOD type cavities. However, there were
Table 13. Statistical analysis comparing magnitude of load and positions on intact teeth and occlusal and MOD preparations

<table>
<thead>
<tr>
<th>Load Position</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Strain (µm/m)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Buccal Principal Strain 1</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Load (N)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>50</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>200</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Buccal Principal Strain 2</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>50</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>200</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Buccal Maximum Shear Strain</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>50</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>200</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lingual Principal Strain 1</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>50</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>200</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lingual Principal Strain 2</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>50</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>200</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lingual Maximum Shear Strain</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>50</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>200</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Kruskal-Wallis Test

P = 5%

* = Significant Difference at 5% level
Figure 45. Variation in principal and shear strains with load and position in a natural tooth prepared with an MOD cavity.
Figure 46. Mean principal and shear strains and angles for a geometric composite molar model subjected to a load of 100 Newtons in a range of positions
Figure 47. Mean principal and shear stresses on teeth prepared with MOD cavities and subjected to an intercuspal load, calculated using a range of values for Young's modulus.
also significant differences in the maximum lingual shear strains between intact teeth and those containing occlusal preparations and between teeth prepared with occlusal and MOD type preparations.

The relationships between loading amplitude and position were compared for intact teeth and those prepared with occlusal and MOD cavities. A non-parametric two-way analysis of variance (Kruskal-Wallis; PROC NPAR1WAY)\(^{(27)}\) was used to compare principal and shear strains for the three different specimen types at the five loads and five positions. The significant differences at the 95% level are indicated in Table 13.

Variations in principal and shear strains over a range of loading amplitudes and at different loading positions are illustrated for one specimen which had been prepared with an MOD cavity in Figure 45. This also suggests a near-linear relationship between applied load and measured strains.

The complex geometry of a natural tooth or molariform composite specimen made it difficult to interpret the relationship between loading position and the magnitude and direction of the resulting principal strains. This was made clearer by the use of more regular geometric composite specimens. Figure 46. illustrates the mean principal and shear strains and associated angles for unrestored specimens having an MOD cavity and subjected to a load of 100N in a range of positions.

The maximum and minimum principal stresses, \(B_{a1}, B_{a2}, L_{a1}, L_{a2}\), were calculated for natural teeth prepared with MOD cavities and loaded intercuspally at a range of loads from the principal strains using equations 5.2 and 5.3. Three different values were substituted for Young's modulus of enamel and the resultant stresses are plotted graphically in Figure 47. Prepared natural teeth and composite specimens were restored using a range of techniques and materials outlined in Figure 36.

A generalised linear modelling procedure (PROC GLM)\(^{(27)}\) was used to evaluate the influence of restoration techniques and loading positions on principal and shear strains for occlusal and MOD specimens subjected to a load of 100N. There were no interactions between restoration techniques and loading positions. With the exception of \(B_{e2}\), principal and shear strains were significantly different at the 5% level between the different loading positions. However, only the maximum buccal and lingual shear strains (\(B_{\tau}, L_{\tau}\)) were
Figure 48. Mean strains on intact and restored teeth loaded intercuspally to 100 Newtons.
Figure 49. (a) Mean principal strains on intact natural teeth loaded to 100 Newtons in two positions

(b) Mean principal strains on teeth restored with bonded composite restorations loaded to 100 Newtons in two positions
(c) Mean principal strains on natural teeth with MOD preparations loaded to 100 Newtons in two positions

(d) Mean principal strains on natural teeth restored with MOD amalgam restorations loaded to 100 Newtons in two positions
Figure 50. Mean principal and shear strains on natural teeth subjected to different operative procedures and loaded to 100 Newtons in a range of positions.

Mean principal and shear strains and confidence levels at 95% for restoration techniques [1, 11, 14 and 21] are shown in Figure 48. for natural teeth loaded intercuspally [load position 1] to 100N. Although the distribution of strains appears to be very similar for the different types of restoration the magnitude of strains is quite different.

In addition to the intercuspal loading position already described strains were also compared between buccal and lingual cusps for MOD type specimens loaded on the lingual incline of the buccal cusp (position [4]) and the buccal incline of the lingual cusp [3] at 100N. The results are presented graphically in Figure 49.(a), (b), (c) and (d).

Mean principal and shear strains together with confidence levels at 95% for natural teeth subjected to a load of 100N at a number of loading positions [1-5] and restored using a range of techniques and materials are presented graphically in Figure 50.

5.4 Discussion

5.4.1 Discussion of the method

Although criteria for the selection of natural teeth as specimens included average dimensions and a crown form representative of a lower molar tooth inherent variations in geometry made precise and repeatable placement and orientation of strain gauges difficult to achieve between different specimens. It was estimated that variations in placement were not greater than ±300μm and orientation was within 10°. In order to minimise any physical and chemical deterioration following extraction all teeth were refrigerated and stored in deionised water for a maximum period of two weeks prior to use.

Strain gauges on three specimens failed and had to be replaced. This was due to fracture of the leads at the junction of the gauge, fracture where the leads emerged from the embedding medium and overheating of one element resulting in an electrical open
In all other respects the strain gauges performed satisfactorily although one gauge suffered a partial debond and had to be reattached.

A pilot study was performed to determine if a satisfactory adhesive bond could be obtained between the composite material used for the model systems and adhesive materials used for the restorations. Shear testing of composite buttons of different configurations resulted in the following mean values for bond strength (n=5):

i) composite model\(^{(12)}\)/restorative composite\(^{(11)}\) = 10.8MNm\(^{-2}\)
ii) composite model\(^{(12)}\)/glass-ionomer cement\(^{(3)}\) = 4.2MNm\(^{-2}\)

Macroscopic examination of the mode of failure appeared to be mostly cohesive for both specimen types.

Although cutting jigs have been utilised by some workers (McCullock & Smith, 1986) for the preparation of cavities in teeth it was considered that the cutting pattern and cavity form resulting from use of these techniques may not be representative of that seen clinically and cavities were prepared freehand using a standardised technique. Mean dimensions calculated from silicone impressions taken of each cavity and observed under a stereo-microscope\(^{(20)}\) were 3.5mm, bucco-lingually and 4.2mm, occluso-cervically. Mesio-occluso-distal cavities consisted of a channel without proximal boxes as it was thought that their presence would increase the variables and complexity of the stress distribution. It was considered that the moulded composite specimens would reduce some of the variables attributable to geometry and cavity dimensions. In an attempt to reduce variables associated with cavity restoration a fixed mass of material (0.31gm) was used for each specimen and this proved to be a satisfactory method for filling cavities.

5.4.2 Discussion of the results

5.4.2.1 Restorative procedures

Restoration of occlusal and MOD cavities in natural teeth and composite specimens with amalgam resulted in small changes in cuspal strain considered to be within the limits of stochastic errors in the investigation (±10\(\mu\varepsilon\)). During setting a typical amalgam alloy may undergo a rapid contraction immediately after condensation followed by a slower
expansion and then a slow contraction. Overall dimensional changes of $+8\mu\text{m/cm}$ have been quoted by Darvell (1991). Three factors probably contribute to the low strains recorded during setting of amalgam restorations in this investigation;

a) The small dimensional change of the alloy used (34).

b) Strains were only monitored for twenty minutes following condensation because of the significance of amplifier drift over extended periods.

c) As amalgam does not adhere to enamel or dentine shrinkage of the restoration would result in the formation of a contraction gap between the tooth and restoration.

It is widely accepted that the volume of a resin based restorative composite decreases when the material polymerises (Goldman, 1983; Walls, McCabe & Murray 1988; Rees & Jacobsen 1989) and that such shrinkage may result in the transmission of stresses to the remaining tooth structure via an adhesive bond (Jensen & Chan, 1985; Meredith & Setchell, 1987). Methods for measurement of the effects of polymerisation stresses on the remaining tooth structure have been relatively crude and have included the use of dial gauges (Causton, Miller & Sefton, 1985), micrometers (Jensen & Chan, 1985) and LVDT's (Pearson & Hegarty, 1987, 1989) to measure cuspal movement and single element strain gauges to measure cuspal strains (Meredith & Setchell, 1987). More detailed information regarding the principal strains can be obtained by the use of rosette type gauges. It can be seen from Figure 42.(a) that individual elements on buccal and lingual rosettes mounted on a natural tooth prepared with an MOD cavity recorded tensile strains during the placement of an enamel etched and dentine bonded composite restoration. Calculation of the principal strains, however, revealed the presence of a complex multi-axial stress field with compressive strains ($B_{e1}$, $L_{e1}$). This was characteristic for all specimens restored with directly placed composite (11) and glass ionomer (3) - composite (11) restorations. The direction of the principal strains varied slightly between specimens and depended on the increment of composite being placed but was approximately 0 and 90° with respect to element [3] on the buccal and lingual gauges. This finding suggests that polymerisation shrinkage does not cause the cusp to behave as a simple cantilever beam in uniaxial bending along the cavity floor. It is likely that a complex, multi-axial stress field
exists, which, because of the change in geometry of the cusp cross section and variations in
the effective elastic modulus as the proportion of enamel to dentine increases towards the
mesial and distal margins resulted in principal strains which did not lie in the long axis of
the tooth. Measurement of cuspal deflection at the cusp tip or cuspal strain using a single,
axially orientated, strain gauge cannot therefore give an accurate representation of the
state of strain.

This concept can be supported by studying Figure 43.(a), (b) and (c) which are
produced from Table 11. Although there are structural differences and slight variations in
group of the cusp cross section and variations in
dentine as the proportion of enamel to dentine increases towards the
cusps which did not lie in the long axis of
It is clear from Figure 42.(a) and (b) that small changes in strain occur even during
etching the enamel around the cavity margin and priming the dentine surface.
Polymerising the dental adhesive with a visible light source causes a larger change and
subsequent polymerisation of the three increments of composite causes a significant
increase in strains. A decrease in strain was sometimes apparent immediately following
polymerisation of an increment of composite (see Figure 42.(b)). This phenomenon was
evident on both buccal and lingual gauges and may be attributable to stress relaxation in
the tooth or composite or partial bond failure at the tooth/restoration interface.

A number of techniques have been suggested to reduce the polymerisation shrinkage
stresses caused by the placement of adhesive composite restorations. These include
decreasing the volume of composite by use of a lining material or the introduction of pre-
polymerised wedges of composite or glass inserts (Donly et al., 1989). A number of workers have advocated polymerisation of composites extraorally (Wendt, 1987) to maximise the polymer conversion, optimise aesthetics and reduce polymerisation shrinkage clinically by using a thin lute of composite to cement the 'inlay'. Table 12. indicates that there are no significant differences in the strains produced by a 'lined' MOD composite restoration and it can also be seen that polymerisation of the composite lute used to cement an 'indirect' MOD composite restoration results in an increase in principal and shear strains. However, if the restorative procedures overall are ranked according to the strains produced an indirectly placed composite restoration resulted in the lowest strains for restorations using resin based materials.

5.4.2.2 Loading procedures

Clinical assessment of the mechanical behaviour of intact and restored teeth is difficult and has largely been restricted to retrospective studies of fractured teeth (see 1.7.5). In-vitro measurements of the fracture resistance of restored teeth have suffered from limitations regarding load application and specimen mounting. A number of parameters may influence in-vitro analysis of the physical behaviour of restored teeth:

i) Load - Amplitude, waveform and position

ii) Specimen - type of natural tooth or model system

iii) Restoration - technique and material

Specimens were subjected to a range of ramp-like loads at different rates applied to a number of points on the tooth surface. Although these parameters do not replicate in-vivo occlusal contacts it was considered, based on a review of the literature, that important clinical characteristics such as loading rate and amplitude fell within the test range. The mechanical behaviour of teeth resulting from these loads was measured as the magnitude and direction of the principal and shear strains at points on the buccal and lingual surfaces of specimens during the loading cycle and at plateaus for specific loads.

It can be seen from Figure 40. that the rate of change of strains on a prepared natural tooth increased with loading rates but the peak strains developed did not differ significantly. Calculation of the linearity and slope of the load-strain curves in Figure 40. revealed that loading and unloading curves deviated by less than 3% at 0.1KNs⁻¹ and 4%
at 0.05KNs$^{-1}$ suggestive of linear, elastic behaviour. However, strains measured at the lowest loading rate (0.01KNs$^{-1}$) exhibited hysteresis which became more significant at higher loads, suggestive of visco-elastic behaviour. As a result the analysis of measurements of peak strains were made based on loading rates of 0.1KNs$^{-1}$.

It can be seen from Figure 44. that the principal and shear strains appear to be proportional to the applied load. Although confidence intervals for the mean strains at the different loads appeared quite large, differences in the strains were still statistically significant. This variation may be attributable to small differences in the position of the intercuspal contact. It is interesting to note that the principal strains are both compressive on the lingual cusp but $B_{c1}$ is tensile on the buccal cusp. This is indicative of the complex, multiaxial stress field on the tooth surface and may have occurred because two of the three contact points made between the steel bearing and the tooth surface rested on the buccal cusp and this may lead to torsional forces making $B_{c1}$ tensile and giving a principal angle of approximately 120°.

Maximum principal and shear strains were compared for intact and prepared (occlusal and MOD) teeth loaded intercuspally. A generalised modelling analysis revealed that there were significant differences in strains produced by the different loads and also differences in buccal and lingual maximum shear strains for prepared and intact teeth. A Bonferroni t test indicated that $B_{c}$ was greater in prepared (either occlusal or MOD) than intact teeth. $L_{c}$ which was higher than $B_{c}$ could be ranked in the order; MOD, occlusal preparation and intact specimen. Although variations in cavity dimensions must be considered, this suggests that the lingual cusp may be weaker than the buccal cusp and derives benefit from the support of the mesial and distal marginal ridges in an occlusal preparation.

This concept was explored further by looking at the effects of altering the position of the applied load. Although in-vitro studies have used a ball bearing to simulate intercuspal contacts in loading extracted teeth (Watts, 1986; Watts, El Mowafy & Grant, 1984; McCullock & Smith 1986) it is known that in-vivo occlusal contacts are often much more diverse. In addition to an intercuspal contact specimens were also loaded at two points on the buccal [2+4] and lingual cusps [3+5] in an attempt to simulate excursive lateral
occlusal contacts and interferences. The strains resulting from such loads may be identified by reference to Figure 49.(a-d) which illustrates the magnitude and direction of the principal and shear strains for a natural tooth prepared with an MOD cavity and subjected to a load of 100N in the five different positions. A non parametric analysis of variance (Kruskal-Wallis test) was used to determine if there were significant differences in the mean resultant strains in intact and prepared teeth subjected to a range of loads (10-200N) in different positions [1-5]. The results, presented in Table 13. did not indicate large significant differences in the strains measured on the buccal cusp. A load of 200N applied to the two positions on the buccal cusp [2 + 4] resulted in significant differences for $B_{el}$ which suggests that this cusp was relatively stiff. Lingual principal strains however were different for all specimen types at much lower loads with significant levels of compressive strain being produced when the lingual cusp was loaded lingually (positions [1] & [3]). Figure 45. illustrates the resultant strains in a single specimen which had been prepared with an MOD cavity and loaded on five positions on the occlusal surface. It is interesting to note that strains are generally lower when the specimen is loaded buccally and lingually [2 + 5]. In addition unloaded cusps sometimes exhibited large strains (i.e. $L_{e2}$ on [4]) which suggested that the overall stress distribution was complex involving the tooth as a whole.

It is clear that the strains produced in teeth subjected to loads are complex and cannot be resolved by using a single element strain gauge. If such a gauge is not orientated in the direction of one of the principal strains the resulting measurement may be meaningless. It was therefore considered important to establish the orientation of the principal strains in relation to loading position. The complex structure and geometry of a natural tooth make this task almost impossible but use of a simple geometrical composite model of the correct proportions made analysis more practicable. The mean principal and shear strains and associated angles on the buccal and lingual surfaces of geometric composite molar models are shown in Figure 46.. Strains on the lingual cusp were minimal because with the exception of position [1] the cusp was not loaded. If the buccal principal strains, $B_{el}$ and $B_{e2}$, and associated angles, $B_{el}$ and $B_{e2}$, are studied for loading positions [2], [7] and [9] it can be seen that as the load moves mesially $B_{e2}$ changes from a tensile to a compressive
strain (62 - 63με) and vice-versa for B_1 (-42 - 33με). The principal angles also change with respect to element (1). This finding is significant because it shows that it cannot be assumed that the principal strains are in a direct line with the applied load although variations in the position of the applied load will result in a change in direction of the principal strains.

Although description of the reaction of a structure to an applied load in terms of the magnitude and direction of the principal strains is accurate such information is often described in the form of principal stresses. It can be seen from equations 5.2 and 5.3 that the principal stresses are directly proportional to strains and are calculated using values for Poisson's ratio and Young's modulus. The difficulties encountered in accurate measurement of E and τ for human enamel have already been discussed in sections 1.6.1 and 2.4.2 and it was considered that a range of values for these elastic coefficients should be applied to any calculations. Figure 47. illustrates the mean principal stresses on prepared teeth resulting from application of an intercuspal load. Three values of E were applied for enamel representing the extremes of the range. It is interesting to note that with the exception of L_e1 stresses are all relatively low even at the highest modulus. L_e1 appears to be large and its principal angle, L_e1 (≈40°) is in line with the applied load on the lingual cusp suggesting that this cusp may be taking a significant proportion of the load.

It has been suggested (McCullock & Smith, 1986) that restorative materials capable of bonding to enamel and dentine may strengthen the remaining tooth structure when compared with a prepared tooth or one restored with a non-adhesive amalgam restoration. Figure 50. illustrates the mean principal and shear strains for a group of teeth that had been subjected to an intercuspal load of 100N following various preparative and restorative procedures. The relative distribution of strains remains the same throughout the operative procedures but their magnitude on the unprepared teeth appears to be significantly less. It is not surprising that strains on the prepared teeth are similar to those with amalgam restorations as an intercuspal load would apply a compressive force to the cusps against which an amalgam material would offer little resistance. Strains for specimens restored with composite materials appeared slightly lower but not significantly different to amalgam, suggesting that the adhesive composite material does not reinforce
the cusps.

A multiple comparison of means was used to determine if there were significant differences in the strains in teeth when restorations were placed in occlusal and MOD cavities. This revealed differences at the 5% level in the maximum shear strains for both types of cavity. A Bonferroni t test showed that the shear strains on intact teeth were significantly lower than those with occlusal cavities, and those with MOD cavities or amalgam or composite restorations.

Changes in strains were studied when teeth subjected to various restorative procedures were loaded at 100N in two positions [3] and [4] in an attempt to assess if the buccal and lingual cusps differed in their ability to distribute stresses. Figure 49.(a-d) illustrate the mean principal and shear strains for this group of teeth. The pattern of distribution of strains appears to be similar for the different groups exhibiting a linear relationship between the principal and shear strains on the cusp that is subjected to the load. The exception to this is in the specimens restored with composite restorations (Figure 49.(b)) which exhibited a different distribution for the buccal cusp. Overall the strains in the intact teeth appear to be considerably lower than either the prepared or restored groups.

If strains in specimens are compared for all the restorative procedures at the different loading positions (see Figure 50.) it is clear that the lowest strains are found in intact teeth; preparing and restoring an occlusal cavity leads to a rise in measured strains which increases following preparation and placement of MOD restorations. Use of composite materials, placed directly or indirectly does not reduce the principal and shear strains to their pre-operative levels.
CHAPTER 6

IN-VITRO STRESS ANALYSIS OF PREPARED AND RESTORED HUMAN TEETH USING THE FINITE ELEMENT METHOD

6.1 Aims and objectives

6.2 Method

6.2.1 Specimen preparation

6.2.2 Pre-processing

6.2.3 Analysis

6.3 Results

6.4 Discussion of the method and results
6.1 Aims and objectives

Experimental results from thermoelastic and strain gauge measurements have shown that the principal surface strains on intact, prepared and restored teeth were directly proportional to the magnitude of applied loads in the range 10-200N at a rate of 0.1KNs\(^{-1}\). Greater strains were measured in prepared than in intact teeth and these were not significantly reduced by placement of an enamel etched and dentine bonded composite restoration. The failure of an 'adhesive' composite restoration to increase the cuspal stiffness of a prepared tooth leading to a decrease in cuspal strain on loading may occur for a number of reasons; bond failure at the tooth/restoration interface resulting from stresses produced during polymerisation of the composite restoration or during occlusal loading. Alternatively a material having a very low elastic modulus will not distribute stresses effectively to the remaining tooth structure.

This finding was in contradiction to that of Douglas (1985) and Lopez, Leitao and Douglas (1991) who observed an increase in cuspal stiffness in extracted teeth following placement of enamel etched composite restorations and resin inlays. Determination of the stress distribution at the tooth/restoration interface during loading would give an insight into the structural interaction between a restoration and the remaining tooth structure. Such measurements are extremely difficult to carry out experimentally but are well suited to analysis using the finite element method. By using finite element analysis it is possible to produce a two or three dimensional model of a prepared or restored tooth and establish how variations in material properties or loading conditions may influence the resulting stresses or displacements.

The finite element method is an approximate technique capable of reliable results only if the geometry, mechanical properties and load case of a model accurately represent the parameters that describe the actual structure. It was therefore considered important to validate any analyses with experimental results where possible. If it is not possible to establish precise values for some parameters (i.e. values of \(E\) and \(v\) for human enamel) then perturbation of a possible range should indicate their relative importance. Validation can be performed by comparison of the principal strains on the surface of a prepared tooth derived from strain gauge measurements with the strains calculated from a finite element
Figure 51. Schematic representation of the section taken from a restored tooth used to provide the geometry for a finite element model.
mesh that closely represents the experimental model. This enables the effects of variations in loading amplitudes and positions on the principal surface strains to be correlated with the distribution of strains throughout the specimen.

Cavity geometry and dimensions may influence the strains produced in a tooth subjected to a load but the significance of this can be difficult to measure experimentally because of the wide range of variables. However, it can be relatively easy to develop a finite element model of a tooth in which it is possible to vary the size, position and shape of a cavity. Effects of different load positions and the influence of adhesive (i.e. composite) and non-adhesive (i.e. amalgam) tooth/restoration interfaces can also be studied.

6.2 Method

It was considered important that the models developed in this investigation were supported by experimental data wherever possible. Mesh geometry was therefore designed based on sections of restored natural teeth that had been previously used for in-vitro strain gauge measurements.

6.2.1 Specimen preparation

The direction of the principal strains measured by rosette strain gauges positioned on the buccal and lingual surfaces of an intact tooth (specimen (c)) resulting from the application of an intercuspal load of 100N (position [1]) were marked with an indelible marker (see Figure 51.). This procedure was repeated for a load of 100N in position [2] on specimen (f). The precise loading position was also identified on each specimen. Each of the two specimens was then mounted, resting on its mesial surface on a glass microscope slide and embedded in a self-polymerising acrylic resin(5). This procedure was performed to ensure that the strain gauges did not become detached during the subsequent sectioning procedure.

Each specimen was sectioned in the plane of the greatest principal strains (B_{e2} and L_{e2}) using a lubricated diamond saw(6). Sections were prepared so that the directions of the principal buccal (B_{t1}) and lingual (L_{t1}) strains coincided (see Figure 51.). Each slice was positioned on a macro-photographic stage(45) and orientated so that the cut surface lay
parallel to the stage surface. Specimens were illuminated with a halogen light source and photographed\(^{(45)(46)}\). A standard length marker was also photographed which enabled large scale monochrome prints to be produced at a known magnification.

Prints were selected to represent bucco-lingual sections of specimens (c) and (f) along the axes of the greatest principal strains for loading positions [1] and [2] respectively. Each print in turn was mounted on a digitising tablet (type GT12-12B; Genius Corpn., Japan \([47]\)) connected to a micro-computer. Approximately 120 x,y co-ordinates were digitised which, it was considered, accurately represented the boundary geometry and outlined material interfaces (i.e. enamel/dentine, dentine/composite etc.) and loading positions. Coordinates were not used directly to specify node positions in the final mesh, therefore selection of inter-coordinate distances was dictated by specimen geometry (i.e. co-ordinates were positioned closer together where there were curves of small radius).

These co-ordinates were stored as an ASCII file which was imported into a spreadsheet package\(^{\text{(26)}}\). A scaling factor was applied to calibrate the digitised co-ordinates and data was edited to produce a file that could be read directly into a FEA pre-processing package (Mystro, v.10.1; FEA Ltd., England \([48]\)).

### 6.2.2 Pre-processing

Pre- and post-processing of the analyses in this investigation was performed using Mystro\(^{\text{(48)}}\) running on a 486DX, DOS based microcomputer with 4Mb of RAM and a 100Mbyte hardisk\(^{\text{(49)}}\). Model geometry was constructed in a series of steps; the model outline was produced from a series of lines, defined by passing cubic splines through the digitised co-ordinates or points. These lines were then used to define surfaces representing the enamel, dentine and cavities of different sizes.

The DeLauney method (LUSAS Theory Manual; FEA, England) was used to mesh the irregular planar surfaces. This enabled surface mesh density to be controlled by grading the lines comprising each surface. Mesh density was increased in regions of interest; interfaces between tooth and restorative materials, location of strain gauges and regions exhibiting acute changes in geometry. Meshes were constructed using two dimensional plane strain triangular elements (type TPN6) which had six nodes and three Gauss points. Some irregularities were apparent in the meshes which were a consequence of the
Figure 52. (a) Two dimensional plane-strain finite element mesh of a bucco-lingual section of a lower first molar tooth
(b) Material property assignment to finite element mesh
(c) Loading and measurement points on finite element mesh
automatic meshing technique (Figure 52.(a)). Elements were checked for excessive
distortion both visually and numerically. Each analysis was completed by defining and
assigning appropriate material properties, boundary conditions and load cases:

a) **Material properties** - For the purpose of this investigation materials were considered
to be isotropic and exhibit linear elastic behaviour. Values for Young’s modulus and Pois-
son’s ratio assigned to surfaces were defined for all materials using experimental results
where possible. Appropriate surfaces comprising the mesh were designated as the enamel,
dentine, restorative material, pulp and the acrylic mounting block (see Figure 52.(b)). The
surfaces comprising the cavity were not assigned material properties in analyses which
contained a preparation. The dimensions of the cavity preparation were varied by
changing the number of surfaces that formed the cavity; the smallest cavity/restoration
was comprised of surfaces 1 - 3, the medium size, surfaces 1 - 6 and the large, surfaces 1
- 9 (Figure 52.(b)).

b) **Boundary conditions** - These provided the reaction to the forces produced by the
load case. The displacement of a vital tooth in its socket is extremely complex (see section
1.7.2) and it is not possible to accurately model the viscoelastic behaviour of the
periodontal membrane because of limitations of currently available finite element
software. Boundary conditions are commonly applied to a mesh by fixing the positions of
a number of nodes in the x,y and z planes which form the border of the mesh. It was
considered that in this investigation fixing the displacement of the nodes which comprised
the root surface of the specimen would produce a ‘hard’ boundary unrepresentative of an
in-vivo environment. Restraining the freedom of the acrylic block in which the specimens
had been embedded proved to be a satisfactory solution as the fixed nodes were positioned
some distance from the specimen and the acrylic provided a degree of damping. The
selection of appropriate mechanical properties for the embedding medium and a brief
discussion of in-vitro mounting methods are discussed in Appendix VI.

c) **Load cases** - These were designed to simulate the magnitude, direction and position
of experimentally applied loads. All analyses were linear, static and represented peak
values (i.e. 10-200N) applied to specimens in-vitro. Loads were applied as concentrated
point loads to one or more nodes on the mesh boundary in five possible positions (A-E;
Figure 52.(c)) as appropriate. Loads were resolved into their x,y components (e.g. a compressive load of 100N applied to an inner cuspal incline buccally was resolved as -70.7, -70.7N). The intercuspal loading position was taken directly from the experimental model and could be related to the mesh which represented a plane sectioned in the direction of the greatest principal strain when an in-vitro specimen was loaded in this position. Although loads were not applied directly in the same position mesio-distally on the in-vitro specimen as on the FE mesh for the eccentric loading positions (B-E) these positions were considered appropriate to enable the effects of different loading positions to be compared on the same FE mesh.

The magnitude and directions of the principal stresses and strains were recorded on the finite element model at the nodes and elements representing the position of the strain gauges to enable comparisons to be made between numerical and experimental results.

6.2.3 Analysis

A series of analyses were performed on the models described using a range of material properties and load cases:

Analysis 1 The significance of variations in the values of the mechanical properties of enamel and dentine.

Reported values for the elastic coefficients of enamel and dentine vary widely and it was considered important to establish how sensitive the results of a linear static analysis would be to variations in Young's modulus and Poisson's ratio for these materials.

An intercuspal load of 100N was applied to nodes in the occlusal fossa (model [1]). A perturbation was performed by carrying out a series of analyses, substituting different values of Young's modulus and Poisson's ratio for enamel and dentine. Young's modulus for enamel ranged from 60 to 84 GNm$^{-2}$ and for dentine from 12 to 24 GNm$^{-2}$. Poisson's ratio for both materials ranged from 0.2 to 0.4. The areas of the mesh representing the pulp tissue and acrylic block were assigned constant values for Young's modulus of 5.0 and 0.001 GNm$^{-2}$ and a Poisson's ratio of 0.4 and 0.3 respectively.

The maximum and minimum principal stresses were recorded on the specimen surface
at the nodes representing the positions of the buccal and lingual strain gauges on the experimental model.

**Analysis 2** Comparison of the principal stresses in a prepared and restored molar tooth subjected to intercuspal loads in the range 10-200N and validation with experimental results

A series of analyses were performed using a mesh of the same geometry as that in Analysis 1 in which total intercuspal loads of 10, 20, 50, 100 and 200N were applied to nodes in position A (Figure 52.(c)). Material properties based on the results of analysis 1 were assigned to the surfaces comprising the enamel (4,6,7,9,10,11) \( [E = 72\text{GNm}^{-2}, v = 0.3] \) and dentine (3,5,8,12) \( [E = 12\text{GNm}^{-2}, v = 0.35] \). Properties for the acrylic block and pulp tissue were as described previously. No material properties were assigned to surfaces 1,2 and 3 to simulate a cavity and values for Young's modulus of 15GNm\(^{-2}\) and Poisson's ratio of 0.3 based on experimental measurements were assigned to emulate an adhesively bonded composite restoration.

The magnitude and direction of the principal stresses and strains were determined at the nodes representing the sites of strain gauges placed on the buccal and lingual surfaces. This enabled numerical results to be validated with experimental strains measured using gauges. Stresses were also recorded at selected nodes along the interface between enamel or dentine and the composite or cavity margin (Figure 52.(c)). The strains and stresses resulting from numerical analyses of a specimen prepared with a cavity and subjected to intercuspal loads of 50 - 200N were compared with the results obtained from the strain gauges mounted on the original tooth under the same conditions.

**Analysis 3** Investigation of the stress distribution along the tooth/restoration interface in a lower molar restored with a bonded composite restoration

A single load of 200N was applied intercuspally to nodes in position A. Material properties assigned to enamel, dentine, and the composite material were the same as those described in analysis 2 and the properties for the composite restoration were assigned to surfaces 1,2 and 3. The resultant principal stresses were represented graphically as...
Figure 53. Maximum and minimum principal stresses in a buccolingual section of a lower molar tooth restored with a bonded composite restoration and subjected to an intercuspal load of 200 Newtons - derived using the finite element method.
Analysis 4  Investigation of the influence of the mechanical properties of the restorative materials on the principal stresses on a molar tooth subjected to an intercuspal load

It has been shown experimentally that there were no significant differences between the magnitude of the principal strains in prepared teeth and those restored with enamel etched and dentine bonded composite restorations. A possible reason for this is that the 'stiffness' of the restoration is much lower than that of enamel and although there is adhesion at the tooth/restoration interface the composite material does not distribute stresses effectively to the remaining tooth structure. This concept was investigated using a finite element model of the same geometry as that in Analysis 2. A single intercuspal load of 200N was applied in position A. Values for the material properties of enamel and dentine were the same as those in analysis 3. A series of analyses were performed in which Young's modulus of the restorative material which was assigned to surfaces 1, 2 and 3 ranged from 5 to 50GNm⁻².

Analysis 5  Investigation of the effects of different loading positions on the principal stresses in a molar tooth prepared with a cavity or restored with a bonded composite restoration

In-vivo teeth may be subjected to a wide range of contacts in addition to those observed in the intercuspal position. It has been shown experimentally that variations in contact position can significantly alter the magnitude and direction of the principal strains in a tooth. Two models were developed, one, a section of a prepared molar tooth and the other, a tooth restored with a composite material. Material properties and mesh geometry were the same as those used in analysis 2. Five analyses were performed for each of the
two model types in which a 100N load was applied to positions on the buccal or lingual cusp (A - E; Figure 52.(a)) representing a range of possible in-vivo contact positions. A direct comparison between experimental and numerical results was not possible for this analysis because the loading positions for the eccentric contacts were not in the same mesio-distal plane as the intercuspal contacts. In the finite element analysis the contacts have been placed in the same plane i.e. on a single mesh to enable a less complicated comparison to be made between the different loading positions. Maximum and minimum principal stresses were recorded at the nodes where strain gauges had been sited and buccally and lingually at the selected nodes on the tooth/restoration and tooth/cavity interface.

**Analysis 6** Investigation of the influence of cavity and restoration size on the principal stresses produced during loading a lower molar tooth in a number of positions

Using the model and geometry previously described it was possible to three generate three sizes (small [1], medium [2] and large [3]) of cavity and restoration by altering the material properties for specified surfaces. The material properties for enamel, dentine and the restoration assigned to surfaces were the same as those used in analysis 2. The different cavity and restoration sizes incorporated the surfaces described previously. Six analyses were performed in total, three each for the prepared and restored model. The three load cases used represented a 100N load applied to different positions on the specimen, on the buccal [D] and lingual [E] cusps and in the intercuspal position [A]. Results were also compared with those for an intact tooth in which surfaces 1,2,4,6,7 and 9 had been assigned properties of enamel and surfaces 3,5 and 8 properties of dentine. The resultant principal stresses were measured on the outer surfaces buccally and lingually at the positions where strain gauges had been placed.

**6.3 Results**

Each analysis produced two results files; one, an ASCII file contained a listing of the problem with node co-ordinates and element data together with the results given as nodal displacements and element stresses. The other was a binary graphics file which enabled a
<table>
<thead>
<tr>
<th>Analysis</th>
<th>Young's Modulus (GN/m^2)</th>
<th>Poisson's Ratio</th>
<th>Principal Stress Buccal Gauge (GN/m^2)</th>
<th>Principal Stress Lingual Gauge (GN/m^2)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Enamel</td>
<td>Dentine</td>
<td>Enamel</td>
<td>Dentine</td>
</tr>
<tr>
<td>1.01</td>
<td>60.00</td>
<td>12.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.02</td>
<td>72.00</td>
<td>12.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.03</td>
<td>84.00</td>
<td>12.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.04</td>
<td>60.00</td>
<td>18.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.05</td>
<td>72.00</td>
<td>18.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.06</td>
<td>84.00</td>
<td>18.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.07</td>
<td>60.00</td>
<td>24.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.08</td>
<td>72.00</td>
<td>24.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.09</td>
<td>84.00</td>
<td>24.00</td>
<td>0.30</td>
<td>0.35</td>
</tr>
<tr>
<td>1.10</td>
<td>72.00</td>
<td>18.00</td>
<td>0.20</td>
<td>0.20</td>
</tr>
<tr>
<td>1.11</td>
<td>72.00</td>
<td>18.00</td>
<td>0.30</td>
<td>0.20</td>
</tr>
<tr>
<td>1.12</td>
<td>72.00</td>
<td>18.00</td>
<td>0.40</td>
<td>0.20</td>
</tr>
<tr>
<td>1.13</td>
<td>72.00</td>
<td>18.00</td>
<td>0.20</td>
<td>0.30</td>
</tr>
<tr>
<td>1.14</td>
<td>72.00</td>
<td>18.00</td>
<td>0.30</td>
<td>0.30</td>
</tr>
<tr>
<td>1.15</td>
<td>72.00</td>
<td>18.00</td>
<td>0.40</td>
<td>0.30</td>
</tr>
<tr>
<td>1.16</td>
<td>72.00</td>
<td>18.00</td>
<td>0.20</td>
<td>0.40</td>
</tr>
<tr>
<td>1.17</td>
<td>72.00</td>
<td>18.00</td>
<td>0.30</td>
<td>0.40</td>
</tr>
<tr>
<td>1.18</td>
<td>72.00</td>
<td>18.00</td>
<td>0.40</td>
<td>0.40</td>
</tr>
</tbody>
</table>

Table 14. Variations in the principal stresses on the buccal and lingual surfaces of an intact tooth subjected to an intercuspal load of 100N with a range of properties for enamel and dentine
Figure 55. Principal stresses in a molar tooth prepared with an MOD cavity and restored with an adhesive composite material, subjected to an intercuspal load of 10-200 Newtons - derived using the finite element method.
<table>
<thead>
<tr>
<th>Load (N)</th>
<th>10</th>
<th>20</th>
<th>50</th>
<th>100</th>
<th>200</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Gauge Position</td>
<td>Buccal</td>
<td>Lingual</td>
<td>Buccal</td>
<td>Lingual</td>
<td>Buccal</td>
</tr>
<tr>
<td>Strain (um/m)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Finite element Method</td>
<td>-5</td>
<td>-12</td>
<td>-10</td>
<td>-24</td>
<td>-25</td>
</tr>
<tr>
<td>Experimental Method</td>
<td>-1</td>
<td>-7</td>
<td>-5</td>
<td>-31</td>
<td>-10</td>
</tr>
</tbody>
</table>

Table 15. Validation of finite element analysis by comparison with experimental data
Figure 56. Mean - difference plot to compare experimental data derived from strain gauge measurements with results from a finite element analysis.
variety of contour and line plots to be constructed. Values for the maximum and minimum
principal stresses, their direction and the relative displacements of the nodes and elements
of interest were extracted from the ASCII files.

Analysis 1

A total of eighteen analyses were performed using a range of values for the mechanical
properties of enamel and dentine. The analysis number together with the appropriate
values for Young’s modulus and Poisson’s ratio of enamel and dentine are presented in
Table 14. together with the principal stresses derived from the nodes representing strain
gauges placed on the buccal and lingual cusps. The stresses resulted from the application
of an intercuspal load of 100N. There was no difference in the angles of the maximum
principal strains with respect to the x axis for the different analyses.

Analysis 2

A series of analyses were performed to determine the maximum and minimum
principal stresses on a prepared and restored tooth subjected to intercuspal loads varying
in magnitude from 10-200N. Stresses were recorded at the same nodes representing the
sites of gauge placement as those used in Analysis 1. In addition stresses were also
measured at four nodes selected at the buccal and lingual enamel/restoration and
dentine/restorations interfaces. Variations in the stresses with applied loads are
represented graphically in Figure 55. Data points have been joined with straight lines as
the relationship between stress and applied load is linear by definition.

The principal strains for the nodes representing the position of the buccal and lingual
strain gauges in the analyses of the model restored with a composite restoration and
loaded between 10 and 200N are listed together with the principal strains, $\varepsilon_2$ and $\varepsilon_2$,
measured experimentally on the original specimen restored using the same technique and
subjected to the same loading parameters in Table 15 and represented graphically as a
mean-difference plot in Figure 56. This enables experimental and numerical results to be
compared directly.

Analysis 3

Contour plots of the maximum and minimum principal stresses for a model restored
Figure 54. (a) Distribution of the maximum principal stresses along section line 1 along the buccal tooth-restoration interface in the Figure (53)

(b) Distribution of the minimum principal stresses along section line 1 along the buccal tooth-restoration interface in Figure (53)
(c) Distribution of the maximum principal stresses along section line 2 along the lingual tooth-restoration interface in Figure (53)

(d) Distribution of the minimum principal stresses along section line 2 along the lingual tooth-restoration interface in Figure (53)
(e) Distribution of the maximum principal stresses along section line 3 taken through the strain gauges in Figure (53)

(f) Distribution of the minimum principal stresses along section line 3 taken through the strain gauges in Figure (53)
Figure 57. Principal stresses in a buccolingual section of a molar tooth subjected to an intercuspal load of 100 Newtons and restored with an adhesive material of variable elastic modulus - derived using the finite element method
with a composite restoration and subjected to an intercuspal load of 200N are illustrated in Figure 53. These plots also illustrate lines (1,2 and 3) along which the plot of results has been sectioned enabling the distribution of stresses to be plotted along bucco-lingual cross sections of the model [1 + 2] and in close proximity to the tooth/restoration interface [3 + 4]. Distribution of the maximum and minimum principal stresses for the four sections are plotted graphically against distance in Figure 54.(a-f).

Analysis 4

A two dimensional mesh of a bucco-lingual section through a lower molar tooth subjected to an intercuspal load of 200N was used to study the effect of variations in the elastic modulus of the restorative material. The analysis was performed six times using values for Young's modulus of the composite restoration of 5, 7.5, 15, 20, 30, 50 and 100 GNM\(^{-2}\). Maximum and minimum principal stresses were recorded as described previously, on the outer part of the buccal and lingual surface of the tooth and at four nodes forming the tooth restoration interface. The results are plotted graphically as stress vs. elastic modulus of the restorative material in Figure 57 and contour plots of the principal stresses with values for the modulus of the restorative material of 5.0 and 50.0 GNM\(^{-2}\) are illustrated in Figure 58 (a+b). A further analysis was performed to investigate the effect of placement of a low modulus lining material (i.e. a glass ionomer cement\(^{(3)}\)) below a composite restoration. Values for Young's modulus of the lining material (comprising surface 3) and the composite restoration (surface 1 + 2), based on earlier experimental measurements, were 15.0 and 4.0 GNM\(^{-2}\) respectively. The resultant principal stresses are listed in Table 16. together with data taken from an analysis of the same model restored with a composite material throughout (E = 15.0 GNM\(^{-2}\)) to assist in comparison.

Analysis 5

Two meshes, one having material assignments defining a cavity and the other a composite restoration, were each analysed five times by applying a load of 100N at six nodes positioned on the tooth surface. Loading direction was selected to represent that encountered clinically with loads on the outer part of the tooth surface [D+E] being
Figure 58. (a) Contour plot of the maximum and minimum principal stresses in a buccolingual section of a lower molar tooth restored with an adhesive material having a value for Young's modulus of 5 GNm$^{-2}$ - derived using the finite element method.
(b) Contour plot of the maximum and minimum principal stresses in a buccolingual section of a lower molar tooth restored with an adhesive material having a value for Young's modulus of 50 GNm\(^{-2}\) - derived using the finite element method.
<table>
<thead>
<tr>
<th>Restoration Type</th>
<th>Composite</th>
<th>Composite + Glass Ionomer</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Principal Stress</td>
<td>Principal Stress</td>
</tr>
<tr>
<td></td>
<td>GN/m²</td>
<td>GN/m²</td>
</tr>
<tr>
<td></td>
<td>Maximum</td>
<td>Minimum</td>
</tr>
<tr>
<td>Buccal (cervical)</td>
<td>2</td>
<td>-11</td>
</tr>
<tr>
<td>Lingual (cervical)</td>
<td>-3</td>
<td>-15</td>
</tr>
<tr>
<td>Buccal (enamel interface)</td>
<td>18</td>
<td>-140</td>
</tr>
<tr>
<td>Lingual (enamel interface)</td>
<td>30</td>
<td>-136</td>
</tr>
<tr>
<td>Buccal (dentine interface)</td>
<td>4</td>
<td>-26</td>
</tr>
<tr>
<td>Lingual (dentine interface)</td>
<td>11</td>
<td>-23</td>
</tr>
</tbody>
</table>

Table 16. A comparison of the stresses resulting from an intercuspal load of 200N in teeth restored with composite and glass ionomer lined composite restorations - derived using the finite element method
Figure 59. Principal stresses in a molar tooth prepared with an MOD cavity and restored with an adhesive composite material, subjected to a load of 100 Newtons in a range of positions - derived using the finite element method.
directed in towards the centre of the tooth and loads on the inner cuspal incline [A,B+C] directed outwards. The resultant maximum and minimum principal stresses are plotted against loading position for the MOD cavity and composite restoration in Figure 59.

Analysis 6

Principal stresses were recorded on the buccal and lingual surfaces of FE models which were loaded in three positions; intercuspally [A], buccally [D] and lingually [E] with a load of 100N. The assignment of the material properties to the relevant surfaces of the mesh were altered to simulate the presence of cavities and composite restorations of three different sizes. The resultant stresses are plotted against cavity size in Figure 60.

6.4 Discussion of the method and results

Finite element analysis is an approximate numerical method used to determine the distribution of stresses and displacements in a structure subjected to a mechanical load. Linear elastic, static analyses using the finite element method are well proven and in view of current limitations of the technique and a lack of knowledge of the loading parameters to which teeth are subjected in-vivo can be considered to be well suited to analysis of the stress distribution in teeth.

Teeth are amongst the most complex structures to be modelled using the finite element method. It was considered that mesh geometry derived from digitised co-ordinates taken from specimens was an accurate technique for representing the complex external form and interfaces of real teeth. Material properties, specifically Young's modulus and Poisson's ratio, applied by previous users of finite element analysis in dental research have most commonly been taken from studies using quasi-static measurements made approximately thirty years ago or dynamic measurements made almost twenty five years ago. The elastic coefficients of restorative materials may vary depending on the type, batch, mixing or polymerisation technique and storage conditions. It was therefore considered inappropriate to utilise values for physical properties of teeth and restorative materials taken solely from the literature in the numerical analyses in this investigation. Where direct measurement of the properties of a material have not proved to be possible (i.e. Young's modulus for Enamel and Poisson's ratio for enamel and dentine) a series of analyses were performed

240
Figure 60. Principal stresses at nodes on the buccal and lingual surfaces of a 2-D section of a lower molar loaded at 100 Newtons in different positions and prepared with cavities varying in size and restored with a composite material.
using a range of possible values to determine the sensitivity of the results to these variables. A variation in Poisson's ratio from 0.2 to 0.4 resulted in a maximum change in principal stress of ±17% measured at the sites of strain gauges which had been placed on the experimental specimen. This range of stresses is in conflict with a comment by Yettram, Wright and Pickard (1976) that the value of Poisson's ratio does not significantly affect the stress distribution in a two dimensional analysis. It was considered that perturbation below a value of 0.2 would be unrealistic and above a value of 0.4 may result in numerical instability as the elements became incompressible. On the basis of the results of Analysis 1 values of 0.30 and 0.35 for Poisson's ratio of enamel and dentine were selected for use in this investigation. It was considered that the slightly higher value for dentine may reflect the increase in water content and organic material present in this material. There appeared to be a complex relationship between the enamel:dentine modulus ratio and the resulting principal stresses (see Figure 61.); a higher modulus ratio resulting in an increase in stress. The nature of this relationship was not easily deduced and may have been dependent on the region of the mesh where stresses were measured. A value for the enamel:dentine modulus ratio could not be derived from experimental values of Young's modulus derived from hardness measurements in an earlier part of this study because of surface damage to the enamel. A number of workers (Waters, 1980; Yettram, Wright & Pickard 1976, Lehman, 1967) have suggested that enamel and dentine may have a modulus ratio of approximately 6:1 and from the results in Figure 61. this value was considered to be an appropriate baseline for use in this investigation. Values for Young's modulus of enamel and dentine were selected as 72,000 and 12,000 GNm⁻².

A number of workers (Craig, Peyton & Johnson 1961, Duncanson & Korostoff 1975) have observed non-linear behaviour in dentine during loading. However, intact and restored teeth subjected to loads in the range 10-200N in the experimental part of this study exhibited linear-elastic characteristics. Numerical analyses of specimens in this investigation were therefore performed using linear elastic parameters. In view of earlier discussions (see 1.4) regarding the difficulties of accurately modelling non-adhesive material interfaces (i.e. amalgam/tooth) analyses were performed in which elements comprising different materials were united at common nodes. It was considered that this
Figure 61. Relationship between the maximum principal stresses from an intercuspal load of 100 Newtons and the ratio of values for Young's modulus for enamel and dentine.
gave an accurate 'theoretical' representation of an adhesive interface.

The wide range of loading parameters seen in-vivo and their simulation in-vitro have been discussed previously (1.7.4) and it is clear that the loads to which teeth are subjected vary considerably in amplitude, duration and waveform. One of the aims of the experimental part of this investigation was to determine the magnitude and direction of the principal strains resulting from the application of a range of loads to intact and restored teeth in-vitro. Load cases in the numerical part of the study were of the same magnitude as the applied loads in the in-vitro experimental measurements with the intention of illustrating stress distributions at peak loads.

The advantages and disadvantages between modelling structures in two and three dimensions have already been discussed in section 1.4.2.1. The application of three dimensional techniques to the analysis of stress distributions in teeth is attractive because of the ability to study a specimen as a whole rather than in sections. However, considerable practical difficulties still exist in accurately defining the geometry and assigning appropriate material properties to a structure of the complexity of a human tooth. Axisymmetric models, although considerably simpler to produce and more rapid to analyse, cannot be considered to accurately represent the geometry of an irregular structure such as a tooth. Taking into account the origins of geometric data, material properties and loadcases it was considered that the two dimensional plane strain numerical analyses of bucco-lingual sections of lower molar teeth performed in this investigation reflected the distribution of stresses through each section to a reasonable level of accuracy.

Morin et al. (1988) developed a 2-D plane strain finite element model to investigate the stress distribution in a tooth restored with a composite restoration. It was stated that '...the distribution of strain within a tooth subjected to a load is insensitive to the location of the cross-sectional plane except at the ends...' suggesting that a section taken at any plane through a tooth would show similar stresses. The experimental results from this investigation have shown clearly that this is not correct and that there is a complex relationship between specimen geometry, point of load application and region of strain measurement.
Appropriate selection of the type of element used was considered to be an important factor which appears to have rarely been discussed in the literature. A finite element mesh for a structure may be generated from its geometry using a relatively large number of elements having a regular simple geometry (e.g. triangular or quadrilateral) or fewer higher order elements having a more complex shape. Advantages of smaller elements are that their solution and accuracy is often well proven and tested (i.e. triangular TPN6 elements in the LUSAS package comply with NAFEM benchmarks, 1984) and a mesh can be constructed with them using automatic meshing techniques. A mesh composed of fewer more complex elements, however may require less computer time to reach a solution. One of the main drawbacks of mapping with simple elements is that the curved boundaries commonly seen on teeth are approximated by a series of short straight lines (or parabolas if parabolic elements are used) which can lead to geometric errors. Such errors can be reduced by using a larger number of elements mapped to boundary coordinates interpolated as cubic splines. This technique was performed in this study using automatic meshing (DeLauney; LUSAS Theory Manual v.10.1, 1991) with triangular plane strain (type TPN6; LUSAS 10.1, England) elements each having six nodes and three Gauss points and was considered to produce satisfactory results. Use of complex elements can lead to inaccuracies as a result of poor mid-point node positioning or excessive element distortion.

Results from a finite element analysis contains nodal displacements in the x, y and z directions and stresses related to elements. Specific information on stresses varies slightly between different software but typically includes; stresses in the x, y and z planes, shear stresses, maximum and minimum principal stresses, equivalent stresses and often Von Mises stresses. Von Mises equivalent stresses have often been quoted in the results of numerical studies of restored teeth (Peters & Poort, 1983). Their use is questionable in this context as they are designed to give a yield stress based on a criteria which assumes that a material has the same properties in tension and compression and this may not be appropriate for brittle materials such as enamel and dentine. In addition Von Mises stresses do not reveal directional information which could be particularly useful in studying the presence of tensile or compressive stresses at the tooth/restoration interface.
for example. Results of numerical analyses in this investigation were recorded as strains
in the x or y direction ($\varepsilon_x$, $\varepsilon_y$) and the maximum and minimum principal stresses ($\sigma_1$, $\sigma_2$)
with the angle of $\sigma_1$ in relation to the X axis ($\Theta_1$).

The majority of analyses were performed using 2-D meshes to compare differences in
stresses and strains resulting from variations in material properties and loading
magnitudes on a model. Relative accuracy between analyses was considered to be more
important than simulating the exact behaviour of a complex model such as a section of a
tooth.

Validation of numerical analyses was carried out by comparison of the principal strains
recorded on a prepared tooth subjected to a range of intercuspal loads in-vitro with a
numerical model produced from a section of the same specimen taken along the axis of
the minimum principal strains. The results (Table 15. and Figure 56.) illustrated a good
correlation between experimental and numerical strains for lingual cusp measurements but
less so for the buccal cusp. The reason for this is uncertain but may be accounted for in
the experimental model because the intercuspal load is formed by a tripod contact forming
two loading points on the buccal cusp which is not represented on the FE mesh.

Analysis 2 was performed to investigate the relationship between the principal stresses
on the surface and at the tooth/restoration interface in prepared and restored teeth with the
magnitude of an intercuspal load. By definition the relationship between the resultant
stresses and applied load was linear and this was evident in Figure 55. There appeared to
be agreement between the stresses recorded at those nodes on the finite element mesh of
the prepared tooth which represented strain gauge positions on the experimental
specimens and the stresses calculated from experimental strain gauge measurements (c.f.
Figures 47. and 55.). Introduction of a material having the properties of a bonded
composite restoration into the cavity in the numerical model resulted in a decrease in
stresses on the outer tooth surface and at the dentine/restoration interface but there were
significant tensile forces acting on the enamel/restoration interface. These tensile stresses
were acting perpendicular to the enamel/restoration interface and this was considered to
be potentially destructive. It is also clear that high compressive stresses may occur along
the interface if the point of load application is in close proximity to the margin of the

246
restoration as in this model.

In analysis 3 the stress distribution along the tooth/restoration interface was determined by taking sections from contour plots of $\sigma_1$ and $\sigma_2$ for a specimen loaded intercuspally to 200N (see Figure 53). The distribution of stresses appeared to be similar along the buccal and lingual tooth/restoration interfaces although the minimal principal stresses were larger buccally. Peak maximum (tensile) principal stresses occurred in the proximity of the amelodentinal junction (Figure 54. (a) + (c)) but the minimum (compressive) principal stresses decreased with distance from the occlusal surface. The applied load of 200N may be considered to be rather high in comparison to normal functional loads but the resultant tensile stresses could be sufficient to cause failure at the tooth/restoration interface. In addition to buccal and lingual sections running occluso-cervically sections of the contour plots were also made buccolingually. The purpose of this was to ensure that the peaks seen in maximum principal stresses at the interface were not solely due to changes in material properties as the section line failed to follow the curve of the interface.

Results from experimental strain measurements in this investigation showed that placement of an enamel etched and dentine bonded composite restoration in a prepared tooth did not significantly alter the levels of measured strain. It was postulated that this may be due to bond failure at the tooth/restoration interface or because the restorative material had insufficient stiffness to support the remaining tooth structure. This was investigated by performing a series of numerical analyses (4) in which Young's modulus of the restorative material was varied between 5 and 100 GNm$^{-2}$ in a specimen subjected to an intercuspal load. The results, presented graphically in Figure 57. indicated relatively small changes in principal stresses with an increase in the elastic modulus of the restorative material. The most significant changes were an increase in maximum principal stresses and a decrease in minimum principal stresses with increasing modulus at the enamel/restoration interface. This finding suggests that, for a restoration of the size and shape used in the model, the material properties of the restoration influenced the local stresses at the tooth/restoration interface but did not significantly alter the overall stress distribution in the tooth.

A further analysis was performed on the same model to investigate the effect of
placement of a low modulus lining material (4.0 GNm\(^2\)) such as a glass ionomer cement. The results illustrated in Table 15. indicated relatively small changes in principal stresses with introduction of the lining material. There appears to be a slight decrease in minimum principal stresses on the surface of the tooth and this may be significant.

The effects of variations in loading position on the principal stresses were investigated in analysis 5. With reference to Figure 51.(c) it was apparent that intercuspal loads placed in close proximity to the restoration margins [A] caused very high stresses at the enamel/restoration interface and placement of occlusal contacts clinically in such positions should be avoided. Contacts farther up the buccal [B] and lingual [C] slopes produced lower levels of stress. Relatively high tensile stresses were observed at the tooth/restoration interface when loads were applied on the inner aspect of the tooth (positions [A], [B] and [C]). It was also apparent that a tooth containing a bonded composite restoration had lower minimum (compressive) principal stresses at the sites of the strain gauges than a prepared tooth when subjected to loads on positions [B] and [C]. This suggested that the restoration may support the remaining tooth structure increasing cuspal stiffness, a finding that was in conflict with experimental results. Overall the stresses appeared to be lower in the model containing the composite restoration.

Three different sizes of cavity and restoration were compared in analysis 6. A load of 100N was applied intercuspally [A], buccally [D] and lingually [E] and stresses were monitored at the sites of strain gauge placement. There appeared to be a near linear relationship between the principal stresses and the cavity or restoration size (Figure 60.). High compressive stresses were evident for the cavities buccally and linguually in the intercuspal loading position [A] but these became tensile for the buccal cusp in the outer buccal position [D] and for the lingual cusp in the outer lingual position [E]. In the intercuspal position compressive stresses appeared to be significantly greater for the mesh prepared with a cavity than that restored with the composite restoration also suggesting that a composite restoration may increase cuspal stiffness. It was interesting to note that the distribution of the magnitude of stresses was similar for the prepared and restored meshes loaded in the intercuspal position. However this pattern changed for the buccal and lingual loading positions; high tensile stresses which increased with cavity size were
evident for the prepared specimens but the greatest stresses in the restored teeth were compressive and these decreased with an increase in restoration size. This may be accounted for by the progressive replacement of enamel which had a relatively high elastic modulus (72.0 GNm$^{-2}$) with a low modulus composite material (15.0 GNm$^{-2}$) resulting in a decrease in stresses.
CHAPTER 7

CLINICAL IMPLICATIONS, PROPOSED FURTHER WORK & CONCLUSIONS

7.1 Clinical implications

7.1.1 The effects of cavity preparation on teeth subjected to simulated functional loads

7.1.2 The effects of cavity restoration on cuspal strains

7.1.3 Cuspal stresses and strains on restored teeth subjected to simulated occlusal loads in-vitro

7.2 Proposed further work

7.3 Conclusions
7.1 Clinical implications

The structure of a tooth has a major influence on its mechanical behaviour. Yet it is often difficult to extrapolate values for properties derived from simple in-vitro laboratory experiments to the clinical performance of intact and restored teeth. Significant research and effort has been put into studying the physical properties of human enamel and dentine as separate tissues although these materials are almost always intimately combined in nature.

The ratio of Young’s modulus for enamel and dentine based on results taken from an experimental indentation method and information drawn from the literature is approximately 6:1 and this, combined with the relative thickness of the two materials is likely to play a major role in determining the structural behaviour of teeth under loading. Although it is evident clinically that enamel which is not supported by dentine due to the presence of a carious lesion or cavity preparation is brittle and liable to fracture there is relatively little information regarding this phenomenon in the literature. Ultrastructural changes have been observed in carious lesions however, which do indicate that the Young’s modulus of carious dentine is significantly lower than that of sound dentine (Briggs, 1985) and this may lead to a decrease in support for the overlying enamel.

It is also important to consider the physical properties of restorative materials in this context. Study of the mechanical interactions between restorations and the remaining tooth structure have been limited largely to in-vitro measurements of the fracture resistance of teeth, in-vivo measurements being almost impossible to carry out. The clinical significance of such methods is difficult to interpret as there are a large number of variables and the correlation between in-vitro and in-vivo loading parameters is open to question. Measurement of mechanical strains on the surface of teeth is a less destructive and more sensitive technique than determination of fracture resistance and enables the effects of loading at low levels to be studied.

One of the primary aims when restoring teeth must be to minimise the risk of subsequent failure. Possible modes of failure include fracture of the remaining tooth structure, of the restorative material or at the interface between the two. Factors
influencing such failure include the amount of remaining tissue, the nature of the applied load, the type of restoration and the materials used. There are also many complex inter-relationships between these.

7.1.1 The effects of cavity preparation on teeth subjected to simulated functional loads

The size of a prepared cavity is dependent on the extent of the caries, the amount of tissue that needs to be removed in order to gain access to the lesion and structural features which may need to be added to aid retention of the restoration. The extent of a preparation determines the amount of remaining tooth structure and influences its resistance to fracture under applied loads. A prepared tooth will fracture when applied stresses and resultant strains reach a point which will depend on a number of factors:

1) the size and position of cavity and amount of remaining tooth structure
2) the ratio of enamel to dentine in the remaining tooth structure
3) the presence of stress concentrations e.g. cracks in the line angles of the cavity
4) structural interactions between the restoration and the remaining tooth structure
5) the magnitude, location and direction of the applied load
6) fatigue behaviour

It was clear from the results of this investigation that preparation of an occlusal or MOD cavity in intact teeth led to an increase in surface strains on subsequent loading. The tooth structure generally behaved as a linear elastic material, strains being proportional to the magnitude of the applied load. The maximum load applied (200N) was probably greater than a normal in-vivo functional load but did not result in irreversible structural changes or specimen fracture. A full-field stress analysis technique also revealed that cavity preparation causes major changes in the distribution of stresses over the tooth surface on loading.

The highest strains were observed in teeth prepared with MOD cavities and it is likely that the mesial and distal marginal ridges which are retained in an occlusal preparation provide a buttressing effect and a degree of cuspal support. Buccal and lingual cusps in
MOD preparations may be considered to behave as cantilever beams.

Cavity size and position also plays an important role in the levels of strain produced in a tooth under load but the effects can be difficult to determine because of variations in the size and shape of specimens and preparations. Using a numerical stress analysis technique and experimentally derived data it was established that stresses were proportional to the level of the applied load in a model containing cavities of different size. It was not possible however, to establish a simple relationship between the level of stress and the cavity size or amount of remaining tooth structure because of the large number of interdependent variables.

7.1.2 The effects of cavity restoration on cuspal strains

Materials used in the placement of intra-coronal restorations can be divided broadly into four groups: amalgam alloys, composites, ceramics and metal castings. The aim of using such materials is to restore the form and function of the tooth and prevent the recurrence of caries. In addition composite and ceramic materials may also restore the aesthetics of the tooth and this is a major reason for their increasing popularity. The mechanical interaction between an amalgam restoration and the remaining tooth structure is limited because of a lack of adhesion at the material interface although it is possible that this may change with the introduction of adhesion promoters capable of bonding to dentine and amalgam. Changes in cuspal strain measured in this investigation during placement of occlusal and MOD amalgam restorations in prepared teeth were relatively small and probably not significant.

Cuspal strains were also measured during placement and polymerisation of enamel etched\(^{36}\) and dentine bonded\(^{37}\) composite\(^{11}\) restorations. The results indicated that significant strains were produced during polymerisation of the material. Strains increased to reach a peak level as each increment of composite was polymerised and then decreased. The restoration and remaining tooth structure are unlikely to undergo sufficient stress relaxation by flow to account for the decrease in strains recorded (Davidson & De Gee, 1984). Partial bond failure may therefore be a potential cause of the fall in strain observed during polymerisation. Placement and polymerisation of glass-ionomer lined
composite restorations resulted in levels of strain comparable to those seen in directly placed composite restorations and the presence of a lining did not appear to lead to a significant decrease in strains. However, polymerisation of the cement lute in indirectly placed composite restorations resulted in significantly lower strains and did not exhibit the post-cure decrease in strain. This could be attributable to the small volume of material polymerised in-situ in the indirect restoration.

7.1.3 Cuspal stresses and strains on restored teeth subjected to simulated occlusal loads in-vitro

Specimens restored with an amalgam alloy\(^{34}\) exhibited strains of similar magnitude and distribution during loading to those seen in prepared teeth. As amalgam does not bond adhesively to enamel or dentine, tensile stresses produced when specimens were loaded in the intercuspal position would not have been transmitted to the restoration but could have resulted in a marginal opening thus producing similar levels of strain to those seen in a cavity alone. Application of loads to the outer surfaces of the buccal and lingual cusps produced significant tensile stresses at the sites of strain gauge placement but compressive stresses were probably present at the tooth/restoration interface. Therefore it could be speculated that an amalgam restoration may have supported the remaining tooth structure when loads were applied in these positions. It was evident from the in-vitro results however, that strain measurements were not significantly different for prepared or restored teeth loaded in these positions, possibly because the remaining tooth structure was of sufficient thickness and stiffness to minimise stresses transmitted to the restoration.

Surprisingly, the magnitude and distribution of the principal strains in in-vitro specimens restored with etched enamel and dentine bonded composite materials was not significantly different from that in prepared teeth. These experimental results were in contrast to those of a finite element analysis in which there was a significant difference in the resultant stresses between models prepared with a cavity and restored with an adhesive composite material. These results could be explained if partial or complete bond failure had occurred in the experimental specimens between the restoration and remaining tooth structure. Such bond failure may have been the result of stresses at the interface resulting
from polymerisation shrinkage of a composite restoration or produced during occlusal loading.

There was little evidence of bond failure which would have been marked by a rapid decrease in cuspal strain during in-vitro specimen loading although stresses measured at the tooth/restoration interface on a finite element model approached the upper limits of reported bond strengths between composites and etched enamel or bonded dentine. High levels of stress were also observed occlusally and at the tooth/restoration interface when loads were applied close to the restoration margin and it is strongly advocated that such loading conditions are avoided where possible clinically.

Interfacial failure of adhesively bonded composite restorations has been observed clinically by Bayne et al. (1991). These authors considered that this was due to adverse occlusal loading and the findings in this study support this theory. It is possible that bond failure may be initiated during restoration and propagated by occlusal stresses during function. These findings indicated that etched enamel and dentine bonded, intra-coronal composite restorations offer limited support to the remaining tooth structure in function. The use of such materials for the purpose of strengthening the remaining tooth structure cannot be recommended.

The effects of variations in the mechanical properties of a restorative material were investigated using the finite element method to determine a value for Young’s modulus which resulted in the optimum stress distribution in a specimen subjected to functional loads in a range of positions. It was considered that matching the modulus of the restorative material to that of the tissue it was replacing could produce the most favourable stress distribution. This could be achieved using a layer of high modulus material (≈ 72GNm⁻²) to replace the enamel and a lower modulus material (≈ 12GNm⁻²) to replace the dentine. The results of a numerical analysis to investigate the effects of variations in the stresses produced in a tooth under loading indicated that there were only small differences in the magnitudes and distribution of stresses for a range of values of Young’s modulus for the restorative material. A further model simulating a composite restoration with a low modulus lining material exhibited only minor variations in stress distribution in comparison with a model having a purely composite
restoration. The finite element results suggested that the matched modulus model or use of a high modulus restorative material (such as a porcelain inlay) resulted in high levels of tensile stress at the tooth-restoration interface during loading which were considered undesirable. A lower modulus material such as composite however, was less stiff and stresses at the interface were lower. High modulus ceramic materials may offer favourable abrasion resistance and aesthetics but there seem to be few advantages over lower modulus composites regarding their structural behaviour.

Overall, results from in-vitro strain measurements suggested that preparation of a cavity in an intact tooth resulted in an increase in stresses on subsequent loading. Experimental and numerical methods showed that stresses were not reduced significantly by placement of an amalgam or adhesive composite restoration at a range of applied loads and positions on the tooth surface.

7.2 Proposed further work

Measurement of mechanical strains using foil resistance gauges is a sensitive and non-destructive method for determining the reaction of a structure to an applied load and has proved to be a useful technique in investigating the physical behaviour of intact, prepared and restored teeth.

The aim of this investigation was to measure the strains and stresses produced in teeth in-vitro during restoration and subsequent loading at levels and rates which were considered to span a range of parameters seen clinically. The results enabled regions of high stress and potential failure to be identified. It is considered important that strains measured in-vitro are validated, where possible, by in-vivo results. The instrumentation and techniques developed in this investigation may be used clinically for this purpose but additional information is necessary so that the applied load can be related to the magnitude and direction of the principal strains.

The considerable variation in occlusal loading parameters described in the literature reflects difficulties in experimental methods. It is considered that use of load sensitive piezo-polymer films which are available in thicknesses of 10\(\mu\)m or less may overcome some of these problems.
In-vivo teeth are most frequently subjected to intermittent loads of low magnitude and it is possible that failure may occur over a period of time as a result of fatigue. Failure of intact and restored teeth under fatigue conditions may be due to crack propagation as a result of localised microscopic plastic deformation. This process has a much longer time scale than a single static test and may be related to a number of factors including loading and environmental conditions. The fatigue behaviour of intact and restored teeth warrants further investigation. Loading waveforms commonly used in fatigue analysis may be standard forms (e.g. sinewave and ramp) or modelled on the load conditions of a real system. Thus, the information derived from in-vivo load and strain measurements could be applied to a fatigue model system in-vitro.

Restorative materials used in this investigation were considered to be representative of the state of the art at the time of testing. The results showed that measurement of cuspal strains gave an indication of the structural behaviour of teeth during restoration and subsequent loading. This study was limited to intracoronal restorations and it would be valuable if a wider range of restorative procedures and materials were investigated. The possible bond failure at the tooth-restoration interface discussed earlier when using bonded composite materials warrants consideration. Methods for eliminating polymerisation shrinkage in restorative materials or overcoming the resultant stresses, possibly by use of an elastomeric bonding agent, must be considered to be a priority area for further research. Mechanical behaviour of geometric and molariform composite model systems designed in this investigation was found to be similar to that of natural teeth and this is potentially useful in reducing the wide range of variables associated with testing natural teeth. Further development of these models may prove useful in evaluating the mechanical properties of a range of restorative materials.

The thermoelastic stress analysis technique produced some very interesting results and is unique in being able to perform a non-contacting full-field analysis of the stress distribution directly on a specimen surface. Restrictions in the method regarding dynamic load application preclude its use in-vivo although further work to investigate the role of extra-coronal restorations such as composite onlays would be useful.

Finite element analysis is an extremely powerful tool for studying stresses in
structures. However, a tooth is very complex and accurately modelling its geometry and properties is a difficult task. An understanding of the magnitude and distribution of stresses at the tooth-restoration interface is increasingly important with the use of adhesion promoters and luting cements. Unfortunately study of this region is not well suited to experimental methods but further work using a numerical analysis technique could yield useful results. Modelling of material boundaries using finite element techniques is difficult and can introduce significant errors into an analysis however recent developments in joint elements may be of benefit.

7.3 Conclusions

The present study has shown in-vitro that preparation of a cavity in intact teeth results in an increase in the principal strains on the buccal and lingual cusps on subsequent application of a range of loads. The magnitude of these strains was greater for MOD than for occlusal cavities. A full-field stress analysis technique based on measurement of the thermoelastic emission from a specimen surface also revealed significant differences in the magnitude and distribution of stresses between intact and prepared teeth subjected to dynamic loads.

Cuspal strains recorded on intact, prepared and restored teeth were directly proportional to an applied load in the range tested (10-200N) and recovery was largely elastic. Specimen loading was performed at three rates (0.01, 0.05, 0.1 kNs\(^{-1}\)) and although there were no significant differences in the peak levels of strain recorded slight non-linearities were apparent on unloading at the lowest rate.

The effect of a range of loading positions representing intercuspal and excursive occlusal contacts on the magnitude and direction of principal surface strains was investigated. It was found that loading position had a significant effect on the principal strains recorded by buccal and lingual strain gauges.

Standardised geometric and molariform specimens representative of a lower molar tooth prepared with an MOD cavity and fabricated from a composite material\(^{(12)}\) were tested alongside natural teeth in an attempt to reduce some of the variables attributable to inherent biological variation. The structural behaviour of these specimens was comparable
to natural teeth although recorded strains were approximately 30% higher, probably an indication of the low modulus of the composite material in comparison to that of enamel.

The magnitude and distribution of the principal stresses in teeth prepared with cavities of different width and height and subjected to occlusal loads was investigated using the finite element method. Although principal stresses measured in the regions of strain gauge placement on experimental specimens were proportional to cavity size overall this relationship was more complex.

Significant changes in cuspal strain were observed during placement of occlusal and MOD type etched\(^{(36)}\) and bonded\(^{(37)}\) composite\(^{(11)}\) restorations which was attributed to polymerisation stresses produced by the materials. A complex multi-axial stress field produced significant tensile and compressive strains. A decrease in cuspal strains observed immediately following polymerisation may have been due to partial bond failure at the tooth-restoration interface. Changes in strain recorded during specimen restoration were comparable for directly placed and glass-ionomer lined\(^{(3)}\) but lower for indirectly fabricated and cemented\(^{(43)}\) restorations.

Subsequent loading of specimens restored with composite materials at a range of loads (10-200N) in different positions produced changes in the principal cuspal strains which were comparable to those that had been observed in the same specimens with preparations and amalgam restorations. This finding was not supported by the results of a numerical analysis based on the same specimens which indicated higher levels of strain in prepared specimens than those restored with composite materials. The stress distribution on the surface of teeth restored with bonded composite materials measured using a thermoelastic analysis technique was significantly different to that observed on prepared teeth.

The findings of this investigation revealed that the application of a range of simulated occlusal loads to teeth in-vitro resulted in an increase in the principal surface strains following cavity preparation. It was not possible to reduce strains to their preoperative levels following restoration of preparations with a range of materials.
APPENDIX I

APPLICATION OF STRAIN GAUGES TO BIOLOGICAL STRUCTURES

The aim of this appendix is to provide a brief overview of the design and use of resistance strain gauges. Topics such as gauge adhesion, environmental protection and measurements on composites and plastics are dealt with in greater detail because it was considered that these have special relevance in strain gauge applications on biological tissues.

1.1 Electrical Resistance Strain Gauges

The electrical resistance of a wire changes in proportion to the mechanical strain applied to it. The strain sensitivity \( S \) of a metallic alloy may be defined as the resistance change per unit of initial resistance divided by the applied strain:

\[
S = \frac{\Delta R}{R_0} / \epsilon
\]  

(1.1)

Constantan, a copper nickel alloy, is commonly used for the fabrication of strain gauges and is photo-etched as a thin foil in a grid pattern. Advantages of this alloy include, a linear strain sensitivity over a wide range of strain, a high specific resistance (0.49\( \mu \)ohmm\(^{-1} \)) and good thermal stability. By photo-etching onto a carrier material, such as a polyimide or an epoxy resin, gauges as small as 0.2mm can be constructed.

1.2 Strain gauge adhesion

A strain gauge is bonded to a test specimen so that strain may be transmitted without attenuation or modification from the specimen surface to the gauge. An adhesive may influence the apparent gauge factor \( (S_g) \), hysteresis, resistance to stress relaxation, temperature induced drift and insulation resistance.

It has been reported (Perry, 1990; Little et al., 1990) that a strain gauge may cause localised reinforcement when bonded to specimens of low elastic modulus. This will result in an incorrect strain measurement. It was considered that this effect may be of special significance when measuring strains on some dental restorative materials and was
therefore investigated experimentally in chapter 3.

Prior to gauge placement a specimen surface should be dry, roughened slightly and free of grease. There are four types of adhesive that are commonly used to bond strain gauges: epoxy, cellulose nitrate, cyano-acrylate and ceramic cements. Epoxy and cellulose nitrate cements are generally considered to be unsuitable for use in biological applications as they require prolonged curing periods at elevated temperatures. Cyano-acrylate adhesives, however, have a rapid curing period (10 minutes) and low film thickness (approximately 10-20\(\mu\)m).

A uniform adhesive layer between a gauge and test specimen is essential and this may be checked by visual inspection. Tapping the gauge with a pencil eraser will elicit a reading of strain if voids are present in the adhesive layer.

1.3 Environmental protection of gauges

Errors may result in strain measurement if a gauge short-circuits because of the presence of moisture. Temperature fluctuations can also result in an apparent change in strain. Environmental protection of strain gauges is therefore important and may be achieved, in the short term, by application of a layer of wax or thin coat of polyurethane varnish and over longer periods by use of a silicone rubber. Effects of temperature changes may be reduced electrically by use of a Wheatstone bridge circuit in which a 'dummy' unstrained gauge forms one arm of the bridge compensating for temperature changes occurring in the 'active' measurement gauge.

Errors in measurement attributable to temperature can be further reduced by use of strain gauges in which the carrier material and gauge alloy are matched to the thermal expansion characteristics of the specimen. The thermal behaviour of foil resistance gauges bonded to materials exhibiting poor thermal conductivity was considered to be a potential problem in this investigation when small gauges were bonded to materials which were very poor heatsinks. An investigation was therefore carried out to determine the thermal behaviour of rosette strain gauges bonded to substrates which had a low thermal conductivity and this is reported in Appendix II.

1.4 Strain gauge measurements on composites and plastics

Strain gauges are most commonly applied to metals but an increasing number of applications are being found for their use on composites, plastics and even biological
tissues such as enamel, dentine and bone. Such materials may have very different physical and mechanical properties from metals.

Polymer based composites often have a low elastic modulus and high Poisson's ratio which may result in strains of higher than 1%. It has been suggested (Perry, 1990) that correction for the transverse sensitivity of strain gauges may be necessary if Poisson's ratio differs greatly from 0.29.

The thermal behaviour of polymers and composites is also important, coefficients of thermal expansion may be considerably greater than for metals making selection of a suitable gauge backing material and alloy difficult. Thermal conductivity of polymers is often low which will reduce heat dissipation from a gauge and limit the maximum excitation voltage.

Although Perry (1990) advocated the use of single-plane rosettes as opposed to stacked type gauges to minimise reinforcement effects and lower heat dissipation this cannot be recommended when measuring strains on small specimens that have a complex, rapidly changing stress field as individual gauge elements will measure different strains.

1.5 Strain gauge instrumentation

When a foil resistance gauge is subjected to a mechanical strain there is a change in resistance which can be expressed as;

$$\Delta R/R = \frac{S}{g} \epsilon_{xx}$$  \hspace{1cm} (I.2)

where the gauge coincides with the x axis and $\epsilon_{yy} = -0.285 \epsilon_{xx}$.

This change can be measured by a potentiometer or Wheatstone bridge circuit. Calibration of a Wheatstone bridge can be carried out by insertion of a fixed value, high accuracy, shunt resistor in parallel with one arm of the bridge and this may expressed as;

$$\epsilon_c = \frac{R}{S (R_2 + R_c)}$$  \hspace{1cm} (I.3)

where

- $\epsilon_c$ = the calibration strain,
- $R_c$ = the value of the calibration resistor,
- $S_g$ = the gauge factor.
The resistor thus simulates the change in resistance that would occur if a gauge was subjected to a calibration strain ($\varepsilon_c$) of, typically 1000$\mu$e.

Alternatively the change in bridge output voltage can be calibrated by direct comparison with mechanical strain on a test specimen. British standard BS 6888:1988 describes the use of such a technique to determine the gauge factor of strain gauges by loading a beam in four point bending.

1.6 Calculation of principal stresses from strain gauge measurements

A strain gauge measures the average state of strain on a discrete area of a specimen surface. A strain field can be described completely if the maximum ($\varepsilon_1$), minimum ($\varepsilon_2$) and principal shear strains ($\varepsilon_3$) and the direction of $\varepsilon_1$ relative to the x axis (given by the principal angle, $\Theta$) are known. The state of strain can be established by use of a single strain gauge if a specimen is subjected to a uniaxial state of stress and the direction of $\Theta_{xx}$ is known. Often, however a stress field is complex and the principal directions are unknown. It is therefore necessary to measure three strains in order to deduce $\varepsilon_1$, $\varepsilon_2$ and $\Theta_{xx}$. This can be carried out using multiple element 'rosette' type gauges which have fixed inter-element angles.

Calculation of the magnitude and directions of the principal strains from individual gauge readings is carried out by the solution of a series of simultaneous equations. Often a number of such calculations need to be performed and a computer program can be written to do this efficiently. Appendix IV describes such a program and discusses the relevant literature.
II.1 Introduction

Although foil resistance gauges have been used successfully to measure strains on biological tissues such as bone (Hylander, 1986) and dental enamel (Jensen & Chan, 1985) their application does present particular problems. These include protection against a hostile environment, obtaining a good adhesive bond between the gauge and substrate and adequate heat dissipation.

Satisfactory environmental protection (against high humidity, moisture etc.) may be achieved by the use of waxes, varnishes or sealing rubbers. Cyano-acrylate adhesives offer a rapid and effective means of bonding strain gauges and have the advantage that they cure at room temperature.

Although ambient temperature variations may be no more than a few degrees the operating temperature of the gauge is also influenced by the power dissipated in the gauge due to the applied excitation voltage. Power dissipated as heat is dependent on the applied voltage and gauge resistance;

\[ P = \frac{E^2}{R} \]  

where

- \( P \) = Power
- \( E \) = Excitation voltage
- \( R \) = Gauge resistance

A number of factors determine the heat dissipation from a strain gauge including:

a) Strain gauge grid area
b) Number and position of gauge elements
c) Type and thickness of base material
d) Type and thickness of adhesive
e) Nature and local dimensions of substrate material
f) Thermal diffusivity of substrate material
g) Nature of environmental protection
h) Temperature and airflow of air over gauge surface

Temperature variations can significantly influence the measurement of strain recorded by gauges. Changes in strain may be due to a change in the strain sensitivity ($S_A$) of the metal alloy used to fabricate the gauge grid, elongation or contraction of the gauge base and grid or a change in resistance due to the influence of temperature on the coefficient of resistivity of the gauge. In practice the variation of strain sensitivity will be negligible if the ambient temperature changes are less than $\pm 10^\circ$C as $S_A/T$ for the copper-nickel alloy (Constantan) commonly used in gauge construction is $0.00735\%/^\circ$ Celsius. If there is a significant difference in the coefficient of thermal expansion between the gauge grid and the base or substrate material there may be a differential expansion between the gauge and substrate due to a temperature change. Temperature compensated gauges are available with thermal characteristics which can be selected to closely match the properties of the substrate material. An apparent strain caused by temperature dependent changes in the resistivity of the gauge material may also be significant and can be described as:

$$\epsilon_a = \left(\frac{\tau_b}{\tau_g}\right) \Delta T \quad (II.2)$$

where

- $\epsilon_a$ = Apparent strain
- $\tau_b$ = Coefficient of thermal expansion of base material
- $\tau_g$ = Coefficient of thermal expansion of grid material

Significant temperature rises as a result of self heating of the strain gauge may also lead to local changes to the material properties of the specimen under the gauge. In practice these potential problems can be ignored because gauges are often bonded to metal substrates which have a high thermal conductivity and are able to dissipate the heat effectively. Achieving an optimum power density ($Pd < 0.0015\ W/mm^2$) may be more
Figure II.1  Position of thermocouples and strain gauges on test specimen
difficult however, when using small gauges bonded to biological tissues such as enamel and bone which are poor thermal conductors.

A large number of factors may influence the operating temperature and hence apparent strain of a strain gauge system and it was considered necessary to measure temperature changes on and within a specimen fabricated from a material having thermal properties comparable to those of enamel and dentine.

II.2 Method and Materials

A test beam of dimensions 50mm x 5mm x 2mm was fabricated from a visible light curing composite material\(^{(11)}\). An unsheathed type K bead thermocouple 0.5mm in diameter was embedded in the geometric centre of the beam at the time of fabrication. Three element, 45° stacked rosette gauges having an element grid area of 0.7mm\(^2\) and gauge resistance of 120Ω ± 0.1Ω\(^{(41)}\) were bonded to each side of the beam directly over the embedded thermocouple (Figure II.1) using a cyanoacrylate adhesive\(^{(21)}\). A further bead thermocouple was placed in mechanical contact with the surface of each gauge. The specimen was mounted in a perspex enclosure on knife edges and remained mechanically unstrained throughout the experiment.

The ambient temperature throughout the investigation was recorded as 21° ± 0.5° and the enclosure prevented temperature changes due to air currents.

The gauges were connected in a half bridge configuration and excited by a variable DC voltage representing increasing power densities for a period of 2 minutes at each voltage. Temperatures was recorded from the thermocouples mounted on each gauge, within the test specimen and for the enclosure using an analogue-to-digital converter connected to a micro-computer at a sampling rate of 0.5Hz .

II.3 Results

The outputs from the thermocouples and amplifiers had been previously calibrated for temperature and linearity over a temperature range of 0-100°C. At no time during the experiments did the temperature recorded by the thermocouple mounted in the enclosure vary by greater than ±0.5°C.
Figure II.2  Surface temperature rise on strain gauge with increasing power density (single gauge)

Figure II.3  Temperature rise inside specimen with increasing power density (single gauge)
Figure II.4 Surface temperature rise on strain gauge with increasing power density (dual gauge)

Figure II.5 Temperature rise inside specimen with increasing power density (dual gauge)
The surface temperature plotted against applied power density in Figure II.2. results from excitation of a single gauge mounted on one side of the specimen. Figure II.3. shows the change in temperature recorded by the thermocouple mounted within the specimen. Figures II.4 and II.5 show the temperature rises recorded on one gauge and within the specimen when gauges on both sides of the specimen are excited simultaneously.

II.4 Discussion

The purpose of these experiments was to record the local temperature changes occurring on strain gauges and within a test material due to self heating of the gauges at a range of power densities. This enables corrections to be made for the apparent strain induced by the effects of temperature related resistance changes and differential expansion between the gauge and substrate material. This was considered to be especially important where small gauges are used and bonded to poor thermal conductors.

The results indicated that the internal temperature rise within a specimen of the described geometry and material type reached 29.1°C when both gauges were excited at 32mWmm\(^{-2}\) (excitation voltage of 2.0V). Such a temperature rise may result in local changes in the material properties of the specimen and Draughn (1981) has shown that an increase in temperature from ambient to 30°C resulted in a significant decrease in the elastic modulus for a similar composite material.

It has been suggested that the optimum power density when measuring static strains using gauges bonded to a substrate which is a poor heatsink (such as fibreglass) is 0.31-0.78 mWmm\(^{-2}\) (Micro-measurements Tech Note TN502, Strain Gauge Excitation Levels). It was found in this investigation that a power density of 2mWmm\(^{-2}\) (0.5V) resulted in a temperature rise of 1.9° and this was considered satisfactory when using strain gauges with a resistance of 120Ω. The advantage of using a higher excitation voltage is to maintain the same output for a lower gain resulting in reduced noise. These could be used with gauges having a higher resistance (320Ω) or by connecting a higher value series resistor in each arm of the bridge.

Figure 6 shows the apparent strain resulting from a change in temperature when there
is a mismatch between the coefficient of thermal expansion of the gauge foil and base and the substrate material which may be expressed as:

$$E_a = (tca - ycb) \Delta T$$  \hspace{1cm} (II.3)

(compensated for mild steel). This effect may be minimised by selecting compensated gauges and minimising any temperature changes.

High excitation voltages may result in significant temperature rises due to self heating effects. Such temperature rises may produce local changes to the material properties of the specimen under test and result in an apparent strain due to a change in resistance and differential thermal expansion between the gauge and substrate. The only accurate way to determine the heat dissipation of a strain gauge system and establish an optimum excitation voltage is by direct measurement of gauge and specimen temperatures in an application. Excitation of the strain gauges in the experimental model at 0.5v (2mWmm$^{-2}$) produced a surface temperature rise of 1.9° C which resulted in an insignificant apparent strain but produced a satisfactory level of gain.
APPENDIX III

STRAIN GAUGE AMPLIFIER DESIGN

III.1 Introduction

In order to be able to carry out the strain measurements in this study and maintain full compatibility with the chosen types of resistance strain gauges it was considered that a strain gauge amplifier with the following specification would be necessary;

Input:
six independent channels (two rosettes)

Bridge configuration:
quarter, half or full bridge configuration with 120 or 350Ω gauges

Bridge supply:
0-10 volts continuously variable

Calibration:
Shunt Calibration

Variable controls/channel:
Bridge supply, Gauge factor (gain), Bridge balance,

Set zero

Output:
±5 volts, low impedance compatible with A-D converter & metered bridge supply and output

Frequency response:
0-1kHz minimum

A number of commercially available strain gauge amplifiers were considered but rejected as being unsuitable because of the lack of a continuously adjustable bridge supply and cost. A low excitation voltage was considered important as the power dissipation of a stacked rosette gauge could be quite high (see Appendix II). It was therefore decided to build a custom design fabricated from commercially available hybrid operational
Figure III.1  Schematic diagram of strain gauge amplifier and unity gain buffer  
(single channel)
amplifiers. Sufficient circuit modifications were made from manufacturers data sheets to warrant a description of the instrument.

III.2 Circuit Description.

The main circuit for each channel is based around a CIL SGA-300 type hybrid strain gauge amplifier supplied in a 24 pin DIL package (Figure III.1). It is important that the common mode rejection factor (CMRR) of the amplifier is high and a 100dB for the SGA 300 is satisfactory. Bridge completion resistors are switchable providing facilities for quarter, half or full bridges at 120 or 350 Ω. All resistors were 1% metal film and the zener diodes were temperature compensated. In the original design it was found that the output noise was too high. This was reduced by connecting a capacitor (C1) in parallel with R1 in the feedback loop which effectively increased the AC CMR and reduced 100mV ripple to 3mV peak-to-peak. However, this was accompanied by a decrease in frequency response. The rise time of the amplifier was tested in the main method and found to be satisfactory. The power supply is a highly smoothed linear, regulated mains supply providing 15-0-15 volts at 2 amps supplying all channels. Full RF and earth leakage filtering is included. The outputs from the SGA 300 are passed through one half of an AD712 high slew rate operational amplifier connected as a unity gain buffer. This is to correct any impedance mismatch between the SGA 300 output and the input of the multiplexer in the A-D converter. Common features to all six channels in the amplifier include meter switching for bridge supply and outputs and incorporation of a shunt calibration resistor in parallel with the active arm of the bridge. The bridge supply voltage was continuously variable and generally used at 0.5V which was low enough to limit heat dissipation from the strain gauges but high enough to provide adequate sensitivity.

III.3 Component List

R1,R3 100kΩ
R2,R6 100Ω
R4 68Ω
R5,R8 10Ω
R7 47Ω
R9 1KΩ
R10,R11 680Ω
All resistors are 1% metal film unless otherwise stated
VR1,VR2 10KΩ
C1,C6,C7 100nF
C2,C5 10nF
C3,C4 10µF
C, 1µF
D1,D2 IN827
T1 BD135
T2 BD136
T3 BC108
IC1 SGA-300 24 pin Strain Gauge Amplifier
IC2 AD712 Operational Amplifier
APPENDIX IV

THE DERIVATION OF PRINCIPAL STRAINS AND ANGLES FROM EXPERIMENTAL STRAIN GAUGE MEASUREMENTS

The large number of experimental measurements of strain made on teeth and model systems using three element, 90°, rosette strain gauges entailed a considerable number of calculations to derive the principal and shear strains and their associated angles (2048 such calculations in Figure 42, alone). The solution of the simultaneous equations necessary to perform this task was well suited to computerisation and a number of programs have been described in the literature to do this. Two such programs (Singh, 1990; Iremonger, 1987) were written in basic (Quick basic v.4.0; Microsoft Corpn., USA) on a PC compatible micro-computer. Running a test file containing sample values of strain from a 90° rosette gauge resulted in numerical (not programming) errors for the resultant principal strains and associated angles when compared with answers calculated manually. One of the problems was that the relative orientation of the strain gauges and angles of principal strains were implied in the results and not stated explicitly. This may be satisfactory when the respective principal strains can be identified from the geometry of the structure and load case but can be difficult on a specimen such as a tooth which has a complex geometry and multi-axial stress field.

A Pascal program based on equation 4.1 was therefore written to calculate principal strains and angles:

$$\varepsilon_\theta = \frac{1}{2}(\varepsilon_x + \varepsilon_y) + \frac{1}{2}(\varepsilon_x - \varepsilon_y)\cos2\Theta + \frac{1}{2}\tau_{xy}\sin2\Theta \quad (4.1)$$

This equation can be applied three times for the three values of $\Theta$ of the rosette gauges. Thus with three known values of $\varepsilon_\theta$ for three known values of $\Theta$, three simultaneous equations will give the unknown strains $\varepsilon_x$, $\varepsilon_y$, and $\tau_{xy}$.

The program was designed to read the raw strains as variables $B_1$, $B_2$ and $B_3$ (or $L$) in an input file and produce an output file containing the variables;

$$B_{e1} \quad \text{maximum principal strain from buccal strain gauge}$$
minimum principal strain from buccal strain gauge
shear strain from buccal strain gauge
the angle of the principal and shear strains \( B_1, B_2, \) and \( B_3 \)
with respect to the angle of element [1] on the buccal strain gauge

Certain combinations of values for \( B_1, B_2, \) and \( B_3 \) which resulted in a division by zero occurred periodically and caused the program to fall over. This was eliminated by writing potential problem data to an exemption file which was then edited by adding a value of 1\( \mu \)
to \( B_1 \) and resubmitted. It was not considered that this procedure would significantly affect the accuracy of the results.

Program Rosette;

\{ Calculate strain data from 45 degree Rosette gauges
Version 2 with error traps for divide by zero. If there is a potential error the input data is written to fexemp.prn.

M.Sherriff 27/9/91 \}

\{Modifications, changing variable names removal of keyboard input etc.
N.Meredith 15/10/91 \}

function epsxy(s1,s2,s3,theta:real): real;
\{ general strain transformation \}
begin
   a1:=2*theta;
   epsxy:=0.5*(s1+s3)+0.5*(s1-s3)*cos(a1)+s2*sin(a1)/2;
end;

function gmax(s1,s2,s3:real): real;
\{ maximum shear strain \}
begin
\[ g_{\text{max}} = 2 \times \sqrt{\\frac{(s_1-s_3)}{2}} \times \left(\\frac{(s_1-s_3)}{2} + \frac{s_2}{2} \right) \]

end;

var
\begin{align*}
B_1, B_2, B_3, ex, ey, gxy, \theta_1, \theta_2, \theta_3, \theta_{\text{amss}}, cr, B_{\text{rs}}, B_{\text{ei}}, B_{\text{et}}, \text{ang} : \text{real}; \\
\end{align*}

\begin{align*}
B_{ga}, B_{ga2}, \theta, ds_1, ds_2, B_{\text{e}_1}, B_{\text{e}_2}, \text{maxss}, B_{\gamma} : \text{real}; \\
\end{align*}

infile, outfile: string;

switch: char;

fin, fout, fex: text;

count: integer;

label 1;

begin

\begin{align*}
\text{count: } &= 0; \\
\text{cr: } &= \frac{180}{\pi}; \quad \{ \text{degree tp radian transform} \} \\
\end{align*}

\{ set switch for output \}

writeln('Do you want output to screen? y or n'); readln(switch);

writeln('Enter the name of the input file'); readln(infile);

assign(fin, infile);

reset(fin);

assign(fout, outfile);

rewrite(fout);

assign(fex, 'fexemp.prn');

rewrite(fex);

while not eof(fin) do

\begin{align*}
\text{begin} \\
\text{count: } &= \text{count} + 1; \\
\text{readln(fin, } B_1, B_2, B_3; \} \\
\end{align*}

\{ readln($B_1, B_2, B_3$); for keyboard input \}

\{ For 45 degree rosette $ex = B_1, ey = B_3$ \}

\{ First data trap \}

if ($\text{abs}(0.0 - (B_1 - B_3)) < 0.00001$) then begin
writeln(fex,count,B_1,B_2,B_3); goto 1;end;

ex:=B_1;ey:=B_3;

{Calculate gxy}
gxy:=2*(B_2-0.5*(ex+ey)); { only valid for 45 degree gauge }
if (abs(0.0-gxy)<0.0001) then begin;
writeln(fex,count:9,B_1:8:2,B_2:8:2,B_3:8:2); goto 1;end;

{ Calculate Principal Plane/Strain }

{principal planes}
ang:=(gxy/(ex-ey));
theta1:=0.5*arctan(ang);
B_{t1}:=theta1*cr;B_{t2}:=B_{t1}+90;

{ Calculate principal strains }
theta2:=theta1+pi/2;
B_{e1}:=epsxy(ex,gxy,ey,theta1); { e1 at m1 }
B_{e2}:=epsxy(ex,gxy,ey,theta2); { e2 at m2 }

{ Calculate Shear Plane/Strain }

{ plane for maximum shear strain }
theta3:=0.5*arctan(-1/ang);
B_{g1}:=theta3*cr;B_{g2}:=B_{g1}+90;

{ maximum shear strain }
B_{g}:=gmax(ex,gxy,ey); { maximum shear strain }

{ Data output }
if ((switch = 'y') or (switch = 'Y')) then begin
writeln('ex = ',ex:8:4,' ey = ',ey:8:4,' gxy = ',gxy:8:4);
writeln(' +ve angles are counter clockwise from the x-axis');
writeln('The principal strains are :');
writeln(B_{e1}:8:4,' at an angle of ',B_{t1}:6:2,' degrees and');
writeln(B_{e2}:8:4,' at an angle of ',B_{t2}:6:2,' degrees to the x-axis.');
writeln('The plane of the maximum shear strain is ',B_{g1}:6:2,' and ',B_{g2}:6:2,' degrees');
writeln('and the maximum shear strain is ' ,B :8:4);
end;
writeln(fout,count:9,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,',',B :8:2,');
1: end;
close(fout);
close(fin);
close(fex);
end.
APPENDIX V

MATERIALS

Where materials or instrumentation first appear in the text they are described in full by product or brand name, stock number, manufacturer and country of origin. Thereafter they are referred to by a superscript referring to details which appear in this appendix:

(1) Stresscoat, brittle coating; Measurements Group Inc., USA.

(2) SPATE 9000; Ometron Ltd., London, England

(3) Vitrebond (also Vitrabond), batch no.900425; 3M (UK) Ltd., Loughborough, England.

(4) ABS potting boxes, stock no. 509-024; R.S. Components Ltd, Corby, England.

(5) Duralay resin; Reliance Dental Mfg. Co., Worth, Illinois, USA.

(6) Diamond band saw; Exakt Ltd., England.

(7) Specimen polishing bed, rotary polishing machine, 1μm diamond paste; Struers Ltd., Glasgow, England.


(9) Miniload 2 microhardness tester; E.Leitz (GmbH), Wetzlar, Germany.

(10) Scanning electron microscope type S8; Hitachi Co., Japan.


(13) Baseline, lining material; Dentsply, Weybridge, England.


(15) Silicone oil, stock no. 556-531; R.S. Components Ltd, Corby, England.

(16) Universal mechanical testing machine, type 1150; Instron Ltd., High Wycombe, England.
(17) Linear variable differential transformer, type AG5; Sangamo-Schlumberger, Hastings, England.

(18) Micrometer screw, type M250; Mitutoyo Co., Japan.

(19) Dual element strain gauges, type FLA-1AS/11; TML, Japan.

(20) Stereo microscope; Weiss Gmbh, Germany.

(21) Cyano-acrylate adhesive for strain gauges, type CN; TML, Japan.

(22) Air drying acrylic varnish, type M-coat D; Welwyn strain measurement Ltd.

Basingstoke, England.

(23) Type K, digital thermocouple thermometer, type 51; Fluke Mfg., USA.

(24) Analogue to digital converter card, type PC26-AD; Amplicon Liveline Ltd., Brighton, England.

(25) PC-XT Microcomputer; Comcen, Swansea, England.

(26) Quattro Pro v2.0 spreadsheet software; Borland (UK) Ltd., Reading, England.

(27) SAS, Statistical software v.6.03; SAS, Medmenham, England.


(31) Formatray acrylic resin; Kerr Mfg. Co., Romulus, USA.

(32) Tungsten carbide bur type 331L; Beavers Dental Co., Ontario, Canada.

(33) Automatrix, matrix bands; L.D.Caulk, USA.

(34) Dispersalloy, amalgam alloy; Johnson & Johnson Corpn., USA.

(35) Thermostatic water bath, model 160; Grant Instruments Ltd., Cambridge, England.

(36) 36% Phosphoric acid gel; 3M (UK) Ltd., Loughborough, England.


(39) Articulating foil; GHM Gmbh., Germany.
(41) Three element, rosette type, strain gauges, type FRA-1AS/11; TML, Japan.
(42) Melamine lower first molar tooth; Columbia Dentoform Mfg. Co., USA.
(43) Experimental dual cure luting cement; batch no.911023; 3M (UK) Ltd.,
(44) Paradox v.2.1 database software; Borland (UK) Ltd., Reading, England.
(45) 105mm macro lens, OM2N 35mm camera body, Macro-photo stage; Olympus
(47) Digitising tablet, type GT 12-12B; Genius Corp., Japan.
(48) Mystro, v.10.1, pre-processing & LUSAS analysis software; FEA Ltd.,
(49) 486 DX personal computer; Viglen Ltd., London, England.
(50) Quickbasic, software; Microsoft Corp., USA.
THE INFLUENCE OF MATERIAL PROPERTIES OF EMBEDDING MEDIA IN IN-VITRO LOAD TESTING

A large number of studies have been performed to determine the effects of cavity preparation and restoration on the mechanical behaviour of teeth by measuring their resistance to fracture under applied loads in-vitro. Specimens have commonly been mounted in a plastic material which is allowed to set or polymerise around the roots of specimens prior to testing. There appears to have been little comment in the literature regarding the influence of the mounting methods and mechanical properties of the embedding material used on the mechanical behaviour of teeth under applied loads. The relationship is likely to be a complex one, related to mechanical properties of the embedding media, the mass of material surrounding the tooth and loading parameters including magnitude and rate.

There are a large number of indeterminant variables associated with in-vitro measurements of the fracture resistance of teeth and the embedding media used may play a minor role. It was considered in this investigation however, that the strain measurements made may be sensitive to the compliance of the supporting structures and it was decided that this warranted investigation. Such an analysis was well suited to the finite element method because of the relative ease with which the effects of variations in material properties and geometry of a model could be studied.

A two dimensional mesh of geometric tooth specimen embedded in a suitable media and mounted in a brass ring and aluminium block was produced using a preprocessing package\(^{(48)}\) on a personal computer\(^{(49)}\) (Figure VI.1). An automatic meshing technique was used but graded in regions of interest, specifically the enamel and dentine forming the crown of the tooth. Triangular TPN3 type elements were used to fabricate the mesh and it was assumed that there was continuity at material boundaries. Simulation of the mechanical behaviour of the periodontal ligament is potentially attractive although limitations of currently available software and a lack of knowledge of the precise mechanism of the tissues supporting a tooth limits this at present.
Figure VI.1  Finite element mesh, loading diagram and surface allocation for a specimen embedded in a mounting block.
Fixed restraints in the x,y plane were applied to the aluminium block at its base in a similar way to the support used for the experimental loading rig. Surface numbers (Figure VI.1) 1,2,7 and 8 were assigned material properties similar to those of dentine ($E = 12,000 \text{ GNm}^{-2}, \nu = 0.3$) and surfaces 3 and 9 properties similar to enamel ($E = 72,000 \text{ GNm}^{-2}, \nu = 0.3$).

An axial load of 200N was applied to the cusp tips throughout all the analyses. The maximum and minimum principal stresses were measured at nodes representing the sites of strain gauge placement on the experimental models (A,B Figure VI.1). The main experimental variable were the mechanical properties of the embedding media surrounding the tooth. It is difficult to establish precise values for Young's modulus of the alveolar bone supporting the teeth although Wainwright et al. (1976) have reported values ranging from 0.2-19.0 GNm$^{-2}$ for trabecular and compact bone. In in-vitro loading studies chemically curing acrylics are commonly used as embedding media and a typical value for their elastic modulus is 5.0 GNm$^{-2}$. A range of analyses were therefore performed in which surfaces 4 and 10 representing the embedding media were assigned values for Young's modulus of 2.0, 5.0, 10.0, 15.0, 20.0, 25.0 and 30.0 GNm$^{-2}$ and 0.3 for Poisson's ratio.

The results revealed only slight variations in the distribution of stresses in the crown of the tooth. Values for the principal stresses measured at the nodes in the proximity of the strain gauges varied by less than 5% for the different material properties of the embedding media. It was concluded that, in the context of this analysis, the mechanical properties of the embedding media would not have a major influence on the resultant strains and a chemically curing acrylic material was selected as being suitable$^{31}$ for testing.
REFERENCES

AHHLGREN, J., OWALL, B. (1970)

ANDERSON, D. J. (1953)
A method of recording masticatory loads; J. Dent. Res. 32: 785-789

ANDERSON, D. J. (1956[a])
Measurement of stress in mastication I; J. Dent. Res. 35 664-670

ANDERSON, D. J. (1956[b])
Measurement of stress in mastication II; J. Dent. Res. 35 671-673

A critical evaluation of indentation techniques for measuring fracture toughness I; J. Am. Ceram. Soc. 64: 533-538

Masticatory movements and the resulting force; Archs Oral Biol. 12: 195-202

BAN, S., ANUSAVICE, K. J. (1990)
Influence of test method on failure stress of brittle dental materials; J. Dent. Res. 69: 1791-1799

BARBER, F. E., LEES, S., LOBENE, R. R. (1969)
Ultrasonic pulse-echo measurements in teeth; Archs. Oral Biol. 14: 745-760

BATES, J. F., STAFFORD, G. D., HARRISON, A. (1975[a])
Masticatory function-a review of the literature I; J. Oral Rehab. 2: 281-301

BATES, J. F., STAFFORD, G. D., HARRISON, A. (1975[b])
Masticatory function-a review of the literature II; J. Oral Rehab. 2: 349-361

Contributing co-variables in clinical trials; J. Dent. Res. 70: 267 (Abstr. No. 14)

BELGEN, M. H. (1968)
Infra-red radiometric stress instrumentation application range study; NASA Report CR 1067, NASA, Washington D.C., USA.

Cuspal failure of MOD restored teeth; Aust. Dent. J. 27: 283-287

Effects of designs of class II preparations on resistance of teeth to fracture; Oper. Dent. 8: 6-10

BONFIELD, W., TULLY, A. (1984)
Thermal properties of bone; J. Biomech. 25: 274-282

BOWEN, R. L. (1956)
Use of epoxy resins in restorative materials; J. Dent. Res 35: 360-369

Tensile strength and modulus of elasticity of tooth structure and several restorative materials; J. Am. Dent. Ass. 64: 378-387
Correlation between strength of bonding to enamel and mechanical properties of dental composites; J. Biomed. Mater. Res. 16: 775-783

BRADEN, M. (1976)

The impact of composite structure on its elastic properties; J. Dent. Res. 65: 648-653

Mechanical properties and filler fraction of dental composites; Dent. Mater. 5: 346-349

Potential for tooth fracture in restorative dentistry; J. Pros. Dent. 45: 411-414

BRIGGS, G. A. D. (1985)

Thermal properties of teeth; J. Dent. Res. 49: 752-755

Modulus of elasticity in bending of composites and amalgams; J. Pros. Dent. 56: 243-248

BRITISH STANDARD 5447. (1977)

BRITISH STANDARD 6888. (1988)
Methods for calibration of bonded electrical resistance strain gauges; British Standards Institution; London, England

BRITISH STANDARD 5199. (1989)
Specification for resin based dental filling materials (Class B); British Standards Institution; London, England

Fracture resistance of premolar teeth restored with MOD composite inlays; J. Dent. Res. 69: 956 Abstr. 14

CAMERON, C.E. (1964)
Cracked tooth syndrome; J. Am. Dent. Assoc. 68: 405-411

CAUSTON, B.E., MILLER, B., SEFTON, J. (1985)
The deformation of cusps by bonded posterior composite restorations: An in-vitro study; Br. Dent. J. 159: 397-400


COOLIDGE, E.D. (1937)

Comparison of three- and four-point flexural strength testing of denture base
polymers; Dent. Mater. 5: 2-5

CHUNG, K.H. (1990)
The relationship between composition and properties of posterior resin composites; J. Dent. Res. 69: 852-856


The microhardness of enamel and dentine; J. Dent. Res. 37: 661-668

CRAIG, R.G., PEYTON, F.A. (1958)[b]
Elastic and mechanical properties of human dentine; J. Dent. Res. 37: 710-718

Measurement of stresses in fixed bridge restorations using a brittle coating technique; J. Dent. Res. 44: 756-762

Measurement of strains in fixed bridges with electronic strain gauges; J. Dent. Res. 46: 615-619

Compressive properties of enamel, dental cements and gold; J. Dent. Res. 40: 936-940

DALLY, J.W., RILEY. (1985)

DARVELL, B. (1990)
Review: Uniaxial compression tests and the validity of indirect tensile strength; J. Mat. Sci. 25: 757-780

DARVELL, B. (1991)

Relaxation of polymerisation contraction stresses by flow in dental composites; J. Dent. Res. 63: 146-149

Microhardness of sound, decalcified and etched tooth enamel related to the calcium content; Caries Res. 8: 135-144

DAVIDSON, G.B. (1964)
A method for three dimensional photoelastic stress analysis of the remaining tooth structure associated with various cavity preparations; J. Dent. Res. 43: p. 912 (Abstr. No. M4A)

Fractography of a Bis-GMS resin; J. Dent. Res. 68: 1194-1198
Development of an artificial oral environment for testing dental materials; Bi-axial force and movement control; J. Dent. Res. 62: 32-36

An artificial oral environment for testing dental materials; IEEE Transactions on Biomedical Engineering 38: 1-7


Dynamic mechanical properties of an inlay composite; J. Dent. 17: 140-144

An in-vitro investigation of the effects of glass inserts on the effective composite resin polymerisation shrinkage; J. Dent. Res. 68: 1234-1237


DOUGLAS, W. H. (1985)

Effects of temperature on mechanical properties of composite dental restorative materials; J. Biomed. Mat. Res. 15: 489-495

A comparative study of stress distribution in human bone under simple and complex loading conditions and as modified by the insertion of a metallic prosthesis; S. P. I. E. Stress and Vibration: Recent Developments in Industrial Measurement and Analysis 184: 111-114

DUNCAN, J. L., NICOL, A. C. (1985)

DUNCANSON, M. G. Jr., KOROSTOFF, E. (1975)
Compressive viscoelastic properties of human dentine; J. Dent. Res. 54: 1207-1212

EAKLE, W. S. (1986)
Fracture resistance of teeth restored with class II bonded composite resin; J. Dent. Res. 65: 149-153

ELDERTON, R. J. (1976)
The prevalence of failure of restorations; J. Dent. 4: 207-210

EL MOWAFY, O. M., WATTS, D. C. (1986)
Fracture toughness of human dentin; J. Dent. Res. 65: 677-681

EL MOWAFY, O. M., WATTS, D. C. (1989)
Stress and deflections in the floor of model cavity preparations; J. Oral. Rehab. 1: 207-215

Effects of cement bases on the stresses in amalgam restorations; J. Dent. Res. 54: 10-15


FERRACANE, J.L. (1985)
Correlation between hardness and degree of conversion during the setting reaction of unfilled dental restorative resins; Dent. Mater. 1: 11-14

Resistance to cusp fracture in class II prepared and restored premolars; J. Pros. Dent. 55: 184-185

Chewing movements in relation to border movements at the first molar; J. Prosthet. Dent. 46: 308-322

Elastic properties of bovine dentine and enamel; Archs. Oral Biol. 15: 787-796

The effect of some components on the rigidity of mandibular bilateral free end saddle dentures; J. Oral Rehab. 7: 423-433

Clinical implications of the response of enamel and dentine to masticatory loads; J. Prosthet. Dent. 64: 446-454

GOLDMAN, M. (1983)
Polymerisation shrinkage of resin based restorative materials; Aust. Dent. J. 28: 156-161

A method for measurement of occlusal forces in three directions; Helv. Odont. Acta 18 7-11

GRANATH, L.E. (1963)
Photoelastic studies on certain factors influencing the relation between cavity and restoration; Odontologisk. Revy. 14: 278-289

GRANATH, L.E. (1964)[a]
Photoelastic studies on occlusal-proximal sections of class II restorations; Odontologisk. Revy. 15: 169-183

GRANATH, L.E. (1964)[b]
Further photoelastic studies on the relations between the cavity and the occlusal portion of class II restorations; Odontologisk. Revy. 15: 290-297

GRIMALDI, J.R., HOOD, J.A.A. (1973)
Lateral deformation of the tooth crown under axial cuspal loading; J. Dent. Res. 52: p.584 (Abstr. No.10)

HAINES, D.J. (1968)
Physical properties of human tooth enamel and enamel sheath material under load; J. Biomech. 1: 117-125
HANNAH, C. McD. (1970)
The tensile properties of human enamel and dentine; J. Dent. Res. 50: 690 (Abstr. 113)

In-vivo cusp fracture of endodontically treated premolars restored with MOD amalgam or MOD resin fillings; Dent. Mater. 4: 169-173

In-vivo fracture of endodontically treated posterior teeth with enamel bonded resin; Dent. Traumatol. 6: 218-225

Thermoeelastic stress analysis; Technical File No. 132, Engineering 25: 234-237

Fracture toughness of human enamel; J. Dent. Res. 60: 820-827

HEARN, E. J. (1981[a])

HEARN, E. J. (1981[b])

HENRY, A. J., SWANSON, S. A. V. (1968)
Experimental healing of fractures in the rabbit tibia; J. Biomed. Mat. Res. 2: 212-217

HIATT, W. H. (1973)
Incomplete crown-root fracture in pulpal-periodontal disease; J. Periodontol 44: 369-376

HICKMAN, J., JACOBSEN, P. H. (1990)
Relationship of cuspal flexure to dentine bonding analysed by the finite element method; J. Dent. Res. 69: 956 Abstr. 12

Finite element analysis of dental polymeric restorations; Clinical materials 7: 39-43

HOOKE, R. (1676)

HOWELL, A. H., BRUDEVOLD, F. (1950)
Vertical forces used during chewing of food; J. Dent. Res. 29: 133-137

HOWELL, A. H., MANLY, R. S. (1948)
An electronic strain gauge for measuring oral forces; J. Dent. Res. 27: 705-710

The finite element method - stress analysis and much more; Machine Design 5: 92-99

HYLANDER, W. L. (1986)
In-vivo bone strain as an indicator of masticatory bite force in macaca fascicularis; Archs. Oral Biol. 31: 149-157

IREMONGER, M. J. (1987)
In: Basic stress analysis; 1st Ed. Butterworths, England

JENKINS, G. N. (1978)
In "The physiology and biochemistry of the mouth" pp. 54-113, Blackwell Scientific
Publications, Oxford, England

JENSEN, M.E., CHAN, D.C.N. (1985)

Stress pattern variations in operatively prepared human teeth, studied by three dimensional photoelasticity; J. Dent. Res. 47: 548-558

JORGENSEN, K.D., MATONO, R., SHIMOKOBE, H. (1976)
Deformation of cavities and resin fillings in loaded teeth; Scand. J. Dent. Res. 84: 46-50

Effects of composite restorations on resistance to cuspal fracture in posterior teeth; J. Pros. Dent. 57: 431-435

KELVIN (THOMSON, W.) (1853)

Mechanical properties of human cartilage; Ph.D. Thesis, University of London

KINLOCH, A.J. (1980)

A three dimensional finite element model; Op. Dent. 13: 128-137

Parameters of MOD cavity preparations: a 3-D FEM study, part II; Operative Dentistry 16: 42-54

KORBER, K.H., KORBER, E. (1967)

KNOOP, F., PETERS, G.C., EMERSON, W.N. (1939)

In: Descriptive dental anatomy; University of Minnesota, Minneapolis, USA.

Effect of prepared cavities on the strength of teeth; Oper. Dent. 6: 2-5

Elastic recovery at hardness indentations; J. Mater. Sci. 16: 2745-2752

Load transfer of posts and cores to roots through cements; J. Prosthet. Dent. 62: 298-302

LEES, S. (1968)
Specific acoustic impedance of enamel and dentine; Archs. Oral Biol. 13: 1491-1500

LEES, S., ROLLINS, F.R. (1972)
Anisotropy in hard dental tissues; J. Biomechanics 5: 557-566

LEHMANN, M.L. (1967)
Tensile strength of human dentine; J. Dent. Res. 46: 197-201

Influences on the bulk fracture incidence of amalgam restorations: a 7 year controlled clinical trial; Dent. Mater. 3: 90-93

LINDQUIST, B., RINGQUIST, M. (1973)
Bite force in children with bruxism; Acta Odont. Scand. 31: 255-258

LITTLE, E.G., TOCHER, D., O’DONELL, P. (1990)
Strain gauge reinforcement of plastics; Strain 26: 91-98

The fracture toughness of dental composites. II The environment and temperature dependence of the stress intensification factor (Kinc); J. Oral Rehab. 9: 133-138

Effect of a new resin inlay/onlay restorative material on cuspal reinforcement; Quintessence Int. 22: 641-645

Occlusal forces in prosthetically restored dentitions: a methodological study; J. Oral Rehab. 11: 29-37

LUSAS THEORY MANUAL v. 10.1. (1991)
pp. 2.35; Fea Ltd., England

MAHLER, D.B. (1958)
An analysis of stresses in a dental amalgam restoration; J. Dent. Res. 37: 516-526

Photoelasticity as a research technique for analysing stresses in dental structures; J. Dent. Res. 34: 831-838

Distribution of occlusal contacts in maximum intercuspation; J. Prosthet. Dent. 62: 238-242

MANLY, R.R., VINTON, P. (1951)
A survey of the chewing ability of denture wearers; J. Dent. Res. 30: 314-317


An investigation of test house variability in the mechanical testing of dental materials and the statistical treatment of the results; J. Dent. 18: 90-97

In-vitro studies of cusp reinforcement with adhesive restorative material; Br. Dent. J. 161: 450-452

MEREDITH, N., SETCHELL, D.J. (1987)
In-vitro experimental stress analysis of bonded composite restorations; J. Dent. Res. 67: 647 (Abstr. 57)

MILLER, R.G., Jr. (1981)
In: Simultaneous statistical inference; Pub. Springer-Verlag, New York.

Fracture strength of amalgam restorations in modern class II preparations with proximal retentive grooves; J. Pros. Dent. 32: 564-571

Fracture strength of human teeth with cavity preparations; J. Pros. Dent. 43: 419-422

Cusp reinforcement by the acid etch technique; J. Dent. Res. 63: 1075-1078

Biophysical stress analysis of restored teeth: experimental strain measurement; Dent. Mater. 4: 41-48

Biophysical stress analysis of restored teeth: modelling and analysis; Dent. Mater. 4: 77-84

The telemetered measurement of strain on the surface of a maxillary complete denture in function; J. Engl. Med. 12: 118-126


NAFEM. (1984)
Guidelines to finite element practice; National Agency for Finite Element Methods and Standards section 0.2-1

Elastic properties of dental resin restorative materials; J. Dent. Res. 53: 1121-1126

Functional loading of the dentition during mastication; J. Prosthet. Dent. 62: 218-228

NEUMANN, H. H., DISALVO, N. A. (1957)
Compression of teeth under the load of chewing; J. Dent. Res. 36: 286-290

NOKUBI, T., YAMAGA, T., OKUNO, Y., TSUTSUMI, S., IDA, K. (1977)
Finite element stress analysis of tooth, periodontal membrane and alveolar bone I - two dimensional model with non-linear behaviour; J. Osaka Univ. Dental School 17: 9-22

NYQUIST, G., OWALL, B. (1968)
Masticatory load registrations during function; Odontologisk Revy. 19: 45-54

Dental amalgam: clinical behaviour up to eight years; Oper. Dent. 5: 24-29

Selected curing characteristics of light activated composite resins; Dent. Mater. 1: 48-54

Composites for use in posterior teeth: Mechanical properties tested under dry and wet conditions; J. Biomed. Mat. Res. 20: 261-271

PARFITT, G. J. (1960)
Measurement of the physiological mobility of individual teeth in an axial direction; J. Dent. Res. 39: 608-611

Cusp movement in molar teeth using dentine adhesives and composite filling materials;
Cusp movement of molar teeth with composite filling materials in conventional and modified MOD cavities; Br.Dent.J. 166: 162-165

PERRY,C.C.(1990)
Strain-gage reinforcement effects on low-modulus materials; In: Stress Analysis Handbook Section IID pp. 35-38 Pub. Inst. of Experimental Mechanics

Modifications of cavity design on stresses in a restored molar; J.Dent.Res.63: 1217-1220

Biomechanical stress analysis of the amalgam-tooth interface; J.Dent.Res.62: 358-362

PEYTON,F.A., MAHLER,D.B., HERSHENOV,B.(1952)
Physical properties of dentine; J.Dent.Res. 31: 366-370

The relationship between the mechanisms of implant and tooth support; Oral Sciences Review 5: 3-23

PICTON,D.C.A., WILLS,D.J.(1978)
Viscoelastic properties of the periodontal ligament and mucous membrane; J.Pros.Dent.40: 263-272

RANDOW,K.(1986)
On the functional deformation of extensive fixed partial dentures; Swed.Dent.J.Suppl.34: 28-32

Fracture properties of human enamel and dentine; J.Dent.Res.55: 154-164

Forces required to crack unfilled and filled molar teeth; J.Dent.Res.59: p.351 (Abstr.No.334)

The polymerisation shrinkage of composite resins; Dent.Mater. 5: 41-44

Fracture resistance of teeth restored with class II composite restorations; J.Dent.Res.63: p.276 (Abstr.No.950)

RENSON,C.E., BRADEN, M.(1971)
The experimental deformation of human dentine by indentors; Archs.Oral Biol.16: 563-572

RENSON,C.E., BRADEN, M.(1975)
Experimental determination of the rigidity modulus, Poisson ratio and elastic limit in shear of human dentine; Archs.Oral Biol.20: 43-47

ROARK,R.J.(1985)

ROCCA,E., BEVER,R.(1950)
The thermoelastic effect; Trans.A.I.M.E.188 327-333
ROONEY, J., STEADMAN, P. (1987)

Stress analysis of the human tooth using a three-dimensional finite element model; J. Dent. Res. 62: 82-86

Correlation of parameters used to estimate monomer conversion in a light cured composite; J. Dent. Res. 67: 932-937

Micro-indentation hardness; J. Dent. Res. 40: 1116-126

Independent movement of cusps during occlusal loading; Dent. Mater. 7: 186-190

SCOTT, I., ASH, M. M. (1966)
A six channel intra-oral transmitter for measuring occlusal forces; J. Pros. Dent. 16: 56-59


Influence of experimental interfering occlusal contacts on the activity of the anterior temporal and masseter muscles during submaximal and maximal bite in the intercuspal position; J. Oral. Rehab. 10: 207-214


SIMONSEN, R. J., BAROUCH, E., GLEB, M. (1983)
Cusp fracture resistance from composite resin in class II restorations; J. Dent. Res. 62: p. 254 (Abstr. No. 761)

SINGH, S. (1990)
Computer analysis of principal strains from rosette measurements; Strain 96: 212-216

Influence of water exposure on the tensile strength of composites; J. Dent. Res. 69: 1812-1816

Fracture resistance of teeth with resin-bonded restorations; J. Pros. Dent. 55: 694-698

Intraoral strain gauge measurements on complete dentures; a methodological study; J. Dent. 19: 80-84

Determination of some compressive properties of human enamel and dentine; J. Am. Dent. Assoc. 57: 487-495

Compressive properties of hard tissues and some restorative
materials; J. Am. Dent. Assoc. 60: 746-751

STANLEY, H.R. (1968)

STANLEY, P. , Chan, W.K. (1985)
Quantitative stress analysis by means of the thermoelastic effect; J. Strain Analysis 20: 129-137

Personal communication

The work of fracture and its measurement in metals, ceramics and other materials; J. Mater. Sci. 1: 296-301

TSUTSUMI, S. , NOKUBI, T. , YAMAGA, T. , OKUNO, Y. , IDA, K. , SAKUDA, M. (1977)
Finite element stress analysis of tooth, periodontal membrane and alveolar bone II - Three dimensional elastic model; J. Osaka Univ. Dental School 17 23-34

TYLDELEY, W. R. (1959)
The mechanical properties of human enamel and dentine; Br. Dent. J. 106: 269-277

VALE, W. A. (1959)
Cavity preparation and further thoughts on high speed; Br. Dent. J. 107: 333-342

A study of the interfacial shear and tensile stresses in a restored molar tooth; J. Dent. 16: 286-293

Why is enamel structurally anisotropic?; J. Dent. Res. 70: 455

A critique of bond strength measurements; J. Dent. 17: 61-67

The physical and mechanical properties of anterior and posterior composite restorative materials; Dent. Mater. 5: 365-368

In "Mechanical design in organisms"; pp. 174 ; Edward Arnold, London, England

The polymerisation contraction of visible-light activated composite resins; J. Dent. 16: 177-181

WATERS, N. E. (1965)
The indentation of thin rubber sheets by spherical indentors; Brit. J. Appl. Phys. 16: 557-563

Some mechanical and physical properties of teeth; In The Mechanical Properties of Biological Materials; Eds. Vincent, J. F. V., Curry, J. D. pub. Cambridge University Press

WATTS, D. C. (1986)
In-vitro biomechanics of lower molars with minimum class II composite restorations; J. Dent. 14: 130-134

Temperature dependence of the mechanical properties of human dentine; Proc. In:

Surface hardness development in light cured composites; Dent. Mater. 3: 265-269

WATTS, D.C., EL MOWAFY, O., GRANT, A.A. (1984)
Mechanical properties of composite-restored lower molars; J. Dent. Res. 63: p. 496 (Abstr. No. 56)

WATTS, D.C., EL MOWAFY, O., GRANT, A.A. (1987)
Fracture resistance of lower molars with Class I composite and amalgam restorations; Dent. Mater. 3: 261-264

Surface elastic constants of composite resins from Knoop indentations; J. Dent. Res. 68: 977 (Abstr. No. 883)

WEBBER, J.M.B. (1985)

WEIBULL, W. (1951)
A statistical distribution function of wide applicability; J. Appl. Mech. 18 293-297

Variations of enamel density in sections of human teeth; Archs. Oral Biol. 12: 85-97

WENDT, S.L. (1987)
The effect of heat used as a secondary cure upon the physical properties of three composite resins. I. Diametral strength, compressive strength and marginal dimensional stability; Quintessence Int. 18: 351-356

WHEELER, R.C. (1971)
in "Dental anatomy, physiology and occlusion"; pp. 267-298; W.B.Saunders, St. Louis, USA

WHITING, R., JACOBSEN, P.H. (1980[a])
Dynamic mechanical properties of resin-based filling materials; J. Dent. Res. 59: 55-60

WHITING, R., JACOBSEN, P.H. (1980[b])

Effects of incremental versus bulk fill technique on resistance to cuspal fracture of teeth restored with posterior composites; J. Pros. Dent. 60: 283-287

Orthodontic tooth movement analysed by the finite element method; Biomaterials 5: 347-351

A finite element stress analysis of an endodontically restored tooth; Eng. in Med. 13: 167-173

Finite element stress analysis of restored teeth; Dent. Mater. 3: 200-206
An investigation of the viscoelastic properties of the periodontium in monkeys;

The evaluation of materials: relationships between laboratory investigations and clinical studies;
Op. Dent. 15: 149-155

WORNER, H.K., ANDERSON, M.N. (1944)

Finite element stress analysis of a class I amalgam restoration subjected to setting and thermal expansion;
J. Dent. Res. 57: 715-723

Finite element stress analysis of the crowns of normal and restored teeth; J. Dent. Res. 55: 1004-1011

ZIDAN, O., ASMUSSEN, E., JORGENSEN, K.D. (1980)
Tensile strength of restorative resins; Scand. J. Dent. Res. 88: 285-289