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Biomechanical Models of Endodontic Restorations

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1. Introduction

Endodontic treatment is one of the most widely used techniques in present-day odontology owing to the tendency to save teeth whenever possible. In endodontic therapy, the injured pulp of a tooth (located in the interior of the tooth and containing nerves and other vital tissues) is cleaned out and then the space is disinfected and subsequently filled with restorative material. This process is commonly known as root canal treatment. The devitalised tooth resulting from endodontics, has a different stiffness and resistance as compared to the original tooth and is less resistant as a consequence of the loss of tooth structure (Walton & Torabinejad, 2002). The use of intraradicular posts has extended as a technique to restore teeth that have lost a considerable amount of coronal tooth structure. After removing the pulp, the intraradicular post is introduced into the devitalised root. The post helps to support the final restoration and join it to the root (Christensen, 1998). Fig. 1a shows the typical structure of a tooth endodontically restored with a post. The post is inserted into the devitalised root canal, which has previously been obturated at its apical end with a biocompatible polymer called gutta-percha. Cement is used to bond the post to the root canal and a core is placed over the remaining dentine and the post. Finally, an artificial crown is used to achieve an external appearance that is similar to that of the original tooth. Nowadays most of the posts are prefabricated in a range of different materials and designs (Scotti & Ferrari, 2004). However, before prefabricated post became generalised, cast post and cores were used as a single metal alloy unit (Fig. 1b). Cast post-core systems take longer to make and involve an intermediate laboratory stage in which the retention system is created, which makes the whole process more costly. In comparison, prefabricated posts do not need this intermediate stage, which means that the whole restoration process can be performed in a single visit and is obviously easier and cheaper for the patient (Christensen, 1998). Nonetheless, the adaptation of the prefabricated posts to the root canal may be less accurate (Chan et al., 1993).
As the endodontically restored tooth is composed of materials that are different to those of natural teeth, it is expected to have a different biomechanical response under oral loads. The deformation of the system under flexural, compressive or tension forces could be different and so its mechanical strength under static or fatigue loads. Ideally, it seems interesting that the biomechanical behaviour of the restored system should preferably resemble that of the original tooth as much as possible in order to avoid failure of the repaired tooth or its

antagonists or adjacent teeth. Nevertheless, this is only an ideal for endodontic therapy and actual restored teeth would differ from this ideal depending on the geometry of the remaining tooth structure, the material and geometry of the components used in the restoration and the technique used by the dentist.

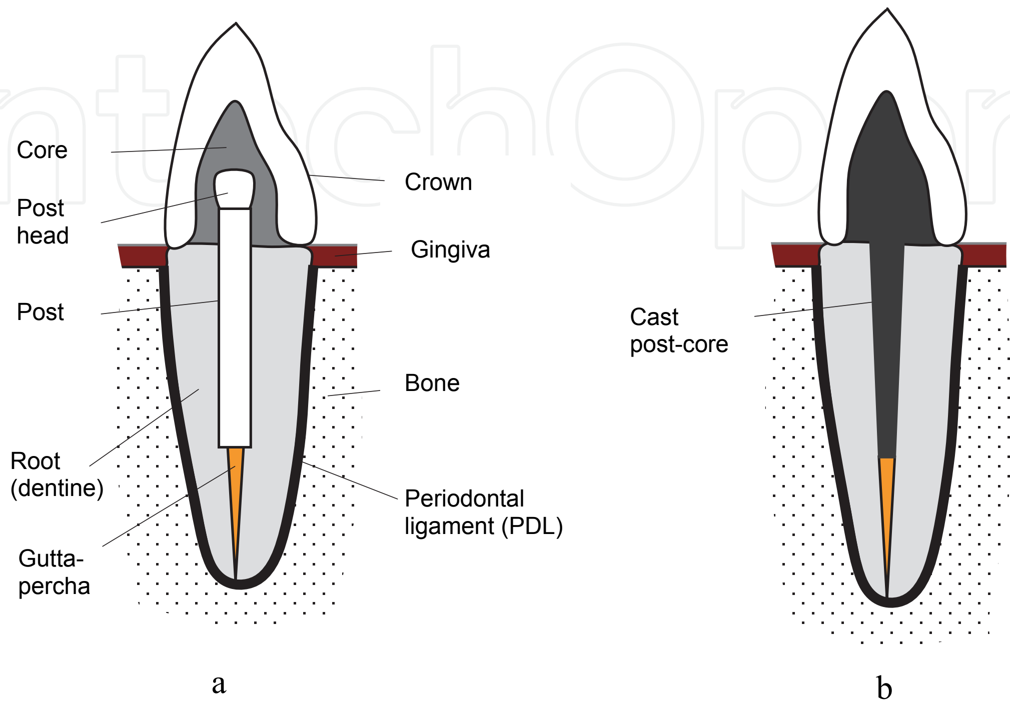

Fig. 1. Endodontic restoration with prefabricated post (a) and cast post-core (b)

The interest in performing an in-depth analysis of the mechanical response of restored teeth becomes apparent from the above considerations and in last few decades a considerable amount of research has been devoted to this objective. Some works have compared this mechanical response with that of natural teeth and others have studied the effect of using different procedures or components for the restoration in order to obtain conclusions about which is preferable for clinical use. The conclusions from these investigations have contributed to the advance of the restorative techniques in Endodontics, that have allowed achieving higher survival rates.

With regard to the methodology used in this field of research, three main lines are distinguished: *in vitro* experimentation, retrospective clinical studies and studies with numerical biomechanical models. The first usually covers the effects of the specific parameters (material, post design, post length or post diameter) on the resistance or the retention of the restoration (Fokkinga et al., 2005; Gallo et al., 2002; Pereira et al., 2006). The majority of these works are static experiments and very few studies have been carried out under cyclic loading or fatigue (Isidor et al., 1999; Sahafi et al., 2005). Retrospective studies are less common given that they require longer times to be performed, often years, and they are based on the study of failures in restorations performed on real patients (Fox et al., 2004; Nothdurft & Pospiech, 2006; Torbjorner et al., 1995). Although working conditions are the current oral dynamic conditions in these studies, the results may be conditioned by loss of control over parameters that are beyond the scope of what is being studied owing to the variability that exists from one patient to another. Finally, the number of studies based on numerical biomechanical models has considerably increased considerably in recent years (Barjau-Escribano et al., 2006; Boschian Pest et al., 2006; Genovese et al., 2005; Pegoretti et al., 2002). These biomechanical models mainly use the finite element analysis (FEA) technique to obtain a numerical representation of the real system and to compute mechanical stresses and strains under simulated loads. In FEA, the system to be studied is divided into a set of small discrete elements (finite elements, FE) defined by a limited number of nodes. A simplified constitutive equation for the finite elements is defined to represent their mechanical response, as a relationship between nodal loads and nodal displacements, and is expressed as the stiffness matrix of the element. By forcing the nodes shared by adjacent elements to have the same displacements, the element stiffness matrices of the elements are combined to give the global stiffness matrix. Nodal displacements can be solved by imposing boundary conditions on this problem, i.e. applied loads and forced displacements and, subsequently, the strains and stresses are then calculated from them. As the size of the finite elements decreases (thus increasing the number of elements), the accuracy of the method increases, but at the same time the computational cost also increases significantly.

The use of FEA for studying the restored tooth has some advantages with respect to the other alternatives cited above. Its main advantage is that it allows a highly controlled analysis of one or several specific parameters on a single tooth model. This results in a better comprehension of either the individual effect or that combined with different parameters. Moreover, numerical analysis with biomechanical models is faster and cheaper than *in vitro* experimentation and eliminates the ethical implications related with the collection of real tooth specimens for *in vitro* experiments or invasive *in vivo* measurements. In contrast, FEA also has some drawbacks namely, the clinical application of their results is conditioned by the accuracy of the model and its previous validation based on experimental data.

In this chapter, a comprehensive review of the state of the art of biomechanical models of endodontic restorations is presented. First, we describe a review of the evolution of these models throughout their brief history. Then, the different aspects related to the definition of the model are analysed with reference to the way they were treated in previous models in the literature. Finally, we provide an overview of the main conclusions reached in previous studies using biomechanical models about the effect of the different parameters involved in the endodontic restoration. Lastly, we present possible lines for future research that can be pursued in this field in order to obtain more comprehensive and accurate models.

2. Historical overview of biomechanical models for simulating endodontic restorations

FEA was developed in the fifties of the last century in the aircraft industry and was not used in dentistry until the seventies. One of the first works on the subject was the doctoral dissertation of Farah, at the University of Michigan, in 1972, which dealt with the simulation of molar restoration using photoelasticity and FEA. Some later works were presented by this research group in following years (Craig & Farah, 1977; Farah et al., 1973; Farah & Craig, 1974). Other pioneering works in those years were presented by Thresher and Saito (Thresher & Saito, 1973) and Selna et al.(Selna et al., 1975). These first works were two-dimensional and were conducted with programs developed by the researchers themselves. Later, the advances in computer resources and in commercial software for FEA extended the use of different commercial programs, allowing more accurate and predictive three-dimensional models to be developed (Asmussen et al., 2005; Barjau-Escribano et al., 2006;

Boschian Pest et al., 2006; Ferrari et al., 2008; Genovese et al., 2005; Gonzalez-Lluch et al., 2009b; Maceri et al., 2009; Rodríguez-Cervantes et al., 2007; Sorrentino et al., 2007).
The number of publications dealing with FE simulation in endodontics has increased exponentially in recent years. For example, in 2011 a manual search in the SCOPUS database in 2011 for papers containing simultaneously the terms *finite element* and *endodontic* within journals on the subjects of Engineering and Medicine and Dentistry found 14 papers until 2000, 31 papers from 2000 to 2009 and 11 papers in the short period including 2010 and first two months of 2011.
The number of elements in the mesh has an influence on the number of degrees of freedom considered in the FE model. Using a mesh with a smaller element size is equivalent to increasing the number of elements and nodes of the model and consequently in the number of degrees of freedom of the problem. This number of degrees of freedom has also increased exponentially in recent years, as shown in Fig. 2.

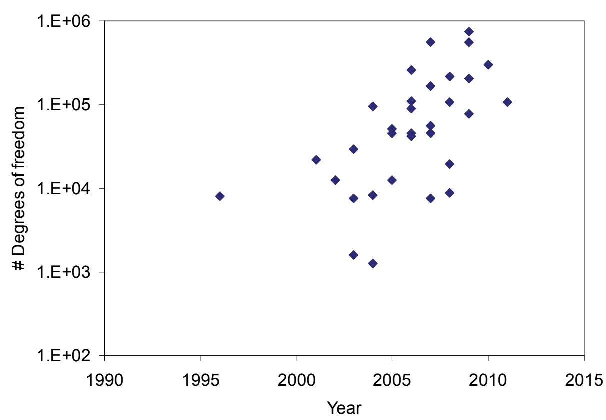

Fig. 2. Number of degrees of freedom of FE models in different works from the literature.

Most of the studies about endodontic restorations using FEA have dealt with incisor teeth, followed by premolars, molars and canines. In the majority of those works, the teeth were analysed under masticatory loads, although other load types, such as bruxism or accidental loads, were also simulated. The development of computers and simulation software has allowed more detailed models. Although an increasing number of studies consider all the components of the restored tooth in the model, other studies still simplify the model by eliminating some of the components. Sometimes the final crown is not included for the sake of simplification (Rodríguez-Cervantes et al., 2007), while in other cases the periodontal ligament (PDL), the luting cement, or both, are left out to simplify the model (Adanir & Belli, 2007; Asmussen et al., 2005; Hsu et al., 2009; Yaman et al., 2004; Zarone et al., 2006).
To date, most of the FE models have been linear and static. In linear static models, the stress-strain curve representing the material behaviour has a constant slope and the materials undergo deformation below their proportional limit. However, in some realistic situations this analysis is not accurate. Wakabayashi et al. (Wakabayashi et al., 2008) cited three main causes of non-linearity: (1) material non-linearities, like the response of the PDL, (2) the changing interrelation of objects, as when loads introduce contacts between components, and (3) geometric nonlinearities, as a consequence of large deformations in some components. In recent years, some works have introduced non-linear FE models as a better way to represent the response of endodontic restorations (Sorrentino et al., 2009; Uddanwadiker et al., 2007). On

the other hand, masticatory loads are variable, implying that the restoration is subject to fatigue, but very few fatigue simulations are found in the literature up until now (Huysmans & Van der Varst, 1993). The authors have recently presented a study on fatigue simulation of a whole endodontic restoration (Sancho-Bru et al., 2009).

3. Biomechanical model definition

The validity of the conclusions reached from biomechanical models of the endodontic restoration will depend on the accuracy with which these models can represent the real system. A number of different factors related to the definition of the model conditions the quality of the results obtained from FEA, namely the accuracy of the geometric representation of the system, the components included in the model, the quality of the discretisation and the constitutive equations for finite elements, the material properties, or the boundary conditions. Moreover, processing and interpreting the results is a challenging task that can compromise accuracy of the conclusions from FEA. All these questions are reviewed in this section.

3.1 Geometric model

The first step in the generation of an FE model is to obtain a geometrical representation of the real system. Some models use a simplified representation of the geometrical shape of the tooth, using elliptical paraboloids or similar shapes to represent the root (Asmussen et al., 2005; Bourauel et al., 2000; Holmes et al., 1996) or cylindrical blocks to represent the alveolar bone (Barjau-Escribano et al., 2006; Holmes et al., 1996; Mezzomo et al., 2011). Other researchers have tried to represent the real geometry better by using anatomical data (Adanir & Belli, 2007; Lanza et al., 2005), X-ray images (Maceri et al., 2007) or CT data (Magne, 2007; Tajima et al., 2009). Authors using two-dimensional data sometimes make use of appropriate algorithms to obtain a three-dimensional model (Maceri et al., 2007). With advances in CAD software and 3D scanning methods, most of the recent works obtain the external geometry of a real representative tooth or a plaster model using 3D digitising scanners (Ausiello et al., 2001; Ferrari et al., 2008; Ichim et al., 2006) and import it into a 3D CAD software application. The geometrical representation of the different components of the restoration is obtained later in CAD using Boolean algebra. The authors recently used this procedure (Gonzalez-Lluch et al., 2009b) to obtain a realistic geometrical model of a maxillary central incisor, which is shown, in a sagittal section, in Fig. 3.

3.2 Components in the model

The natural tooth is the reference for comparing the biomechanical behaviour of an endodontic restoration. Several works in the literature have modelled the natural tooth (Dejak et al., 2003; Middleton et al., 1996; Zarone et al., 2006). Most of modelled natural teeth included enamel, dentine, cortical and cancellous bone, pulp and ligament (Middleton et al., 1996; Rees & Jacobsen, 1997). Some of them did not include the ligament (Zarone et al., 2006). These studies used the model of the natural tooth to compare the calculated biomechanical response with the experimental results (Rees & Jacobsen, 1997), to evaluate the behaviour of a restored tooth against a natural tooth (Davy et al., 1981; Maceri et al., 2009; Soares et al., 2008b; Zarone et al., 2006) or both (Ferrari et al., 2008). Cementum is not considered in most of the models, due to its reduced thickness, although a recent work (Ren et al., 2010) has reported the importance of the cemento-dentinal junction and cementum in stress distribution in the root and the ligament.

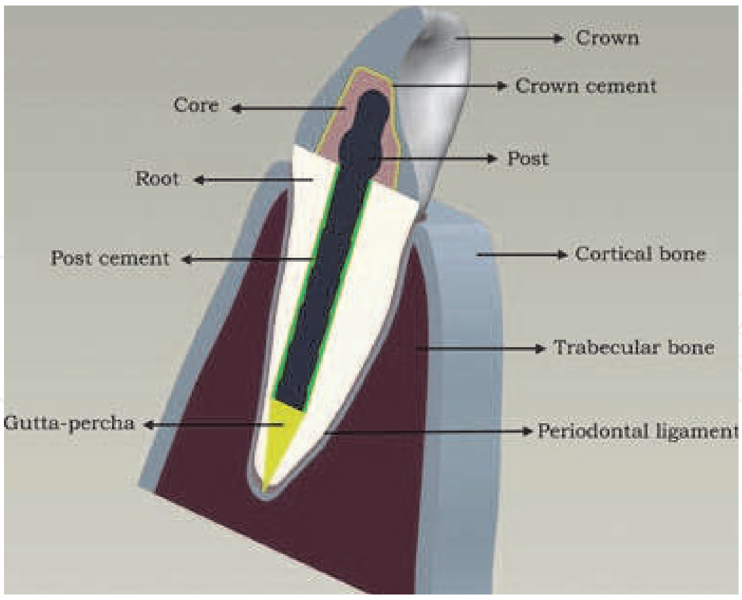

Fig. 3. Sagittal section of an endodontically restored maxillary central incisor

Previous works simulating post-core endodontic restorations have not always included all the actual components of the restoration. Some works assumed that bone deformation is negligible and did not include this component in the model (Boschian Pest et al., 2006; Davy et al., 1981; Maceri et al., 2009). Not all models in the literature include the bone. When included, sometimes it is modelled with uniform properties (Imanishi et al., 2003; Nakamura et al., 2006), although most of the models consider the cancellous bone and a layer of cortical bone near the bone surface (Asmussen et al., 2005; Gonzalez-Lluch et al., 2009b).

Thin structures, such as the PDL and cement are not always considered in the FE models. In natural teeth, the PDL thickness is approximately between 0.125 mm and 0.375 mm (Rees & Jacobsen, 1997). This range is covered in different biomechanical models (Asmussen et al., 2005; Boschian Pest et al., 2006; Ferrari et al., 2008; Ichim et al., 2006; Soares et al., 2010). However, some works did not include the ligament (Yaman et al., 1998; Zarone et al., 2006), despite it has been reported that stress distribution is affected to an important degree by this omission (Davy et al., 1981) or even by a geometrically simplified representation (Toms & Eberhardt, 2003). When cement layers are considered in the biomechanical model, they are usually represented with a constant small thickness (Asmussen et al., 2005; Maceri et al., 2009; Okamoto et al., 2008; Schmitter et al., 2010). However, some works simplify the model by obviating the cement in the interface between the core and the crown (Gonzalez-Lluch et al., 2009b; Lanza et al., 2005) or even in the interface between the post and the root (Adanir & Belli, 2007; Hsu et al., 2009).

Several types of endodontic posts can be found in the literature (Christensen, 1998; Fernandes et al., 2003). Originally, cast post-core systems were used as a single metal alloy unit. Subsequently, prefabricated posts made out of stainless steel, titanium or precious metals were used, while more recently prefabricated fibre posts were introduced. All these post types have been studied with FE models.

Some earlier laboratory studies maintain that preparing a ferrule might improve the resistance of post-core system (Assif & Gorfil, 1994; Aykent et al., 2006; Ichim et al., 2006; Tan et al., 2005; Zhi-Yue & Yu-Xing, 2003). Accordingly, many works in the literature prepared a ferrule of varying heights in the biomechanical model: 2 mm (Mezzomo et al., 2011; Uddanwadiker et al., 2007), 2.5 mm (Coelho et al., 2009) , 1 mm (Toksavul et al., 2006). Ichim et al. (Ichim et al., 2006) prepared biomechanical models without a ferrule and with a ferrule of different heights (0.5, 1.0, 1.5, 2 mm) to study its effect on the strength of the restoration.

3.3 Mesh definition

Mesh definition is a key point in FEA. The mechanical behaviour of a continuous domain with an infinite number of degrees of freedom is approximated in the model with the simplified mechanical response of a set of discrete finite elements with a limited number of degrees of freedom. This process occurs when a mesh of finite elements is defined for the original system. The accuracy of the results is highly dependent on the characteristics of this mesh. A coarse mesh will produce results in a short computation time, but the accuracy of these results will be compromised. A finer mesh may improve the validity of the results but with the cost of a higher processing time. A coarse mesh is affected by two main sources of error: firstly, an error is introduced in representing the boundaries of the real geometry; and secondly, the mechanical response of the continuous material in each finite element is represented by a simplified polynomial function, which is only an approximation of the actual response. In biomechanical applications, such as dentistry, the geometry of the real system is highly irregular, so the mesh density should be enough to obtain a good geometrical representation.

Generation of a good mesh of FE for a restored tooth is always a challenge. The complicated shape of the tooth makes it difficult to produce a mesh manually. Moreover, some elements of a dental restoration present dimensions that are quite different from one another. For example, the luting cement used to bond the post to the root, or the PDL, have a thickness of a fraction of a millimetre, which is much smaller than the post dimensions, although its total length would be of several millimetres. To obtain a good mesh, a small element size has to be used in this thinner component, but such a size cannot be maintained for all the components, because, otherwise, the number of degrees of freedom would increase to a limit beyond the capacity of the computer or the computation time would increase excessively. A good meshing strategy should increase mesh density in areas where a greater expected stress gradient is expected or with thinner components, and increase the size of the elements in other parts of the model. Two different basic strategies can be used to improve finite element results from a first tentative model, namely the h-method or the p-method (Zienkiewicz & Taylor, 1989). In the h-method, the mesh is refined using elements of a smaller size, whereas in the p-method, the size of the elements is maintained but the order of the interpolation function is increased. A mixed strategy that combines both options can also be used. Most of the works in the literature about endodontic simulation use finite elements with linear or quadratic interpolation and refine the mesh using smaller elements to improve the representation, so the h-method is prevalent.

The first FE models used two-dimensional finite elements to represent the tooth and endodontic restoration (Craig & Farah, 1977; Davy et al., 1981), assuming the axisymmetric or plane strain hypothesis. In these planar models triangular or quadrangular elements with three or four nodes, respectively, are used to mesh the system. The size of the finite elements in these first models was relatively large owing to the limitations in computer resources. Davy et al. (Davy et al., 1981) used quadrangular elements for an incisor and tested different meshes to decide on the mesh density by comparing displacements of selected nodes. Finally the finer mesh tested was selected, with elements of about 1 mm in the coronal-apical direction and with a dimension close to 0.3 mm in the lingual-vestibular direction in the cervical area. Middleton et al. (Middleton et al., 1996) used quadrangular elements with eight nodes to study the effect of the PDL in bone modelling. Although the mesh used for tooth and bone was relatively coarse, the ligament was modelled with a much finer mesh, as it was the part of interest in that particular study. More recently, Pegoretti et al. (Pegoretti et al., 2002) also used plain strain two-dimensional elements and modelled the endodontic restoration using about 4000 quadrilateral elements of four nodes. The mesh presented in that work employed elements with an aspect ratio close to one and smaller elements were used near the cement and for the ligament. The aspect ratio for a finite element is the ratio of the longest to the shortest side of the element and should be as close to one as possible in order to minimise possible inaccuracies in the model. In other recent two-dimensional model, convergence tests were used to recommend the element size of 0.1 mm (Coelho et al., 2009). These planar FE models are simplifications of the real system and hence limited. Although they can be explained by the computer limitations in the last few decades, they are no longer justified nowadays. Some authors present pseudo three-dimensional models to represent the tooth (Adanir & Belli, 2007; Li et al., 2006). They obtain a planar representation of the geometry from a sagittal section of the tooth and then a solid from an extrusion normal to this planar model, using a fictitious small thickness. This approach is doubtful because it is as assuming a planar stress hypothesis, which is difficult to justify with the geometry of the real tooth. Similarly, an assumption of planar strain or axisymmetry is difficult to justify if a three-dimensional model can be used.

Three-dimensional models have been used extensively for simulating endodontic treatment since the beginning of the century (Asmussen et al., 2005; Ausiello et al., 2002; Barjau-Escribano et al., 2006; Boschian Pest et al., 2006; Genovese et al., 2005; Lanza et al., 2005). In recent years some of the most complete three-dimensional models ever developed for studying the biomechanics of endodontic restorations have been reported (Ferrari et al., 2008; Garbin et al., 2010; Gonzalez-Lluch et al., 2009b; Maceri et al., 2009; Mezzomo et al., 2011; Okamoto et al., 2008; Schmitter et al., 2010). In these three-dimensional models, tetrahedral or hexahedral finite elements are used. The mesh is created normally using the mesher of a commercial finite element package software, such as Ansys (Garbin et al., 2010; Hsu et al., 2009; Mezzomo et al., 2011), MSC-Nastran (Gonzalez-Lluch et al., 2009b; Maceri et al., 2009; Rodríguez-Cervantes et al., 2007), MSC-Marc (Okada et al., 2008; Pegoretti et al., 2002) or Cosmos (Ichim et al., 2006). In those cases, a tetrahedral mesh is the typical option because of the complicated geometries of the tooth. As an example, Fig. 4 shows a sagittal section of a mesh using tetrahedral elements for a restored incisor, generated by the authors using the Pro/Engineer FEM mesher. Some works, however, have used hexahedral elements (Ferrari et al., 2008; Okamoto et al., 2008; Schmitter et al., 2010; Zarone et al., 2006), which provide models with good results using fewer degrees of freedom, but at the cost of a more difficult mesh generation. In some models, elements with quadratic interpolation, i.e. 20-node hexahedral or 10-node tetrahedral, are used to improve the results (Maceri et al., 2009; Schmitter et al., 2010). Some attempts have been made in recent years to automate the process of creating high-quality meshes using hexahedral elements from anatomical CT data (Clement et al., 2004).

The use of convergence tests is the most commonly reported method for deciding mesh density in the majority of previous works (Ferrari et al., 2008; Maceri et al., 2009; Schmitter et al., 2010). Some authors report special attention to the convergence of results near the more stressed area, such as the post-cement interface and the cemento-enamel junction (Garbin et al., 2010; Zarone et al., 2006). However, no explicit mention is made of the parameter used to test the convergence in most of the works. Hsu et al. (Hsu et al., 2009) based convergence on the total deformation and established a difference of 1% as the limit to consider convergence. The number of nodes and elements for some three-dimensional models of the endodontically restored tooth are presented in the Table 1. It can be observed that a lower number of elements are used when the mesh is composed of hexahedral elements. The average size of elements in most of these models is close to 0.2 mm or 0.3 mm (Gonzalez-Lluch et al., 2009b; Maceri et al., 2009). The authors have shown that a mesh control of 0.3 mm is a good compromise between accuracy and computation time (Gonzalez-Lluch et al., 2009a).

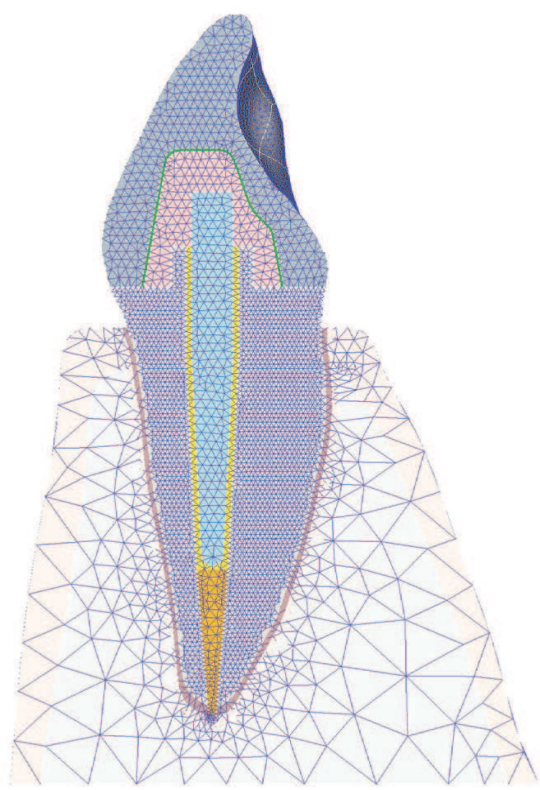

Fig. 4. Sagittal section of a typical mesh with tetrahedral elements for a restored tooth

Reference	# Elements	# Nodes	Type of element
(Zarone et al., 2006)	13272	15152	8-node hexahedral
(Ferrari et al., 2008)	31240	35841	8-node hexahedral
(Gonzalez-Lluch et al., 2009b)	399000	69000	4-node tetrahedral
(Maceri et al., 2009)	130000	185000	10-node tetrahedral
(Schmitter et al., 2010)	95000	100000	20-node hexahedral
(Mezzomo et al., 2011)	23000-33000	30000-42000	4-node tetrahedral

Table 1. Type and number of elements and number of nodes of selected FE models

The assumption of perfect adherence between adjacent components is the rule of thumb for most of the FE models presented in the literature, i.e. nodes located on the contacting surfaces of two components are shared by these components. Post cement is normally supposed to be joined perfectly to the root dentine and to the post. However, some works represent some exceptions to this habitual modelling hypothesis. Ausiello et al. (Ausiello et al., 2002) modelled the adhesive layer in composite restorations of premolars using spring elements, which allow different normal and shear stiffness to be represented, and a good agreement with experimental data was obtained. This representation of cement was also repeated in later works by the same group (Ferrari et al., 2008; Maceri et al., 2009). Schmitter et al. (Schmitter et al., 2010) modelled different debonding states in the adhesive layers, by introducing gaps between components, to investigate possible failure modes and to compare with experimental results and a similar procedure was used by Ichim et al. (Ichim et al., 2006). Asmussen et al. (Asmussen et al., 2005) used contact finite elements to simulate non-bonding states. These contact elements allow a separation between contacting surfaces but not interpenetration between them and can simulate shear friction in the contact area. Finally, in one recent study (Garbin et al., 2010) the cement layer was modelled using shell elements in Ansys.

Different types of finite elements have been used to represent the PDL. Most of the works represent the ligament with solid isotropic elements filling the reduced thickness of the ligament, but other authors use spring elements (Maceri et al., 2009), beam elements (Freitas Junior et al., 2010) or shell elements (Garbin et al., 2010). Finally, in order to represent the viscoelastic behaviour of the ligament better, (Natali et al., 2004) proposed the use of a finite element with customised formulation, using a specific Helmholtz free-energy density. To the authors' knowledge, however, this model has not been used to simulate endodontic restorations.

3.4 Material properties

Material properties are one of the key points in the definition of the model. Accuracy of the results will depend on a good representation of the actual material properties, which are not always well known. Most of the FE models assume the materials of the restored tooth are isotropic, homogeneous, elastic and linear. A material is considered homogenous if its composition can be considered as being the same at all the points, isotropic if the mechanical response is indifferent to the direction of the applied forces, elastic if it returns to the undeformed position after releasing the applied load, and linear if load and deformation are proportional. These four hypotheses are typical in a great part of all FE models in engineering, resulting in linear models for which the response (stress or strain) is proportional to the applied load. Although they allow the definition and interpretation of the model to be simplified to an important extent, these hypotheses can introduce errors in the representation of certain materials. In endodontic restoration, some materials such as bone, dentine, fibre posts or the PDL are known to be anisotropic and even not homogeneous. For example, van-Ruijven et al. (van Ruijven et al., 2006) reported that neglecting the non-homogeneity of the alveolar bone results in important changes in stresses and strains in the model. Moreover, the mechanical response of the PDL has been reported to be non-linear and viscoelastic (Pini et al., 2002). However, not only is the model that is used to represent material important, but also the numerical values of the mechanical properties, such as Young's modulus and Poisson's ratio, are of great importance to obtain an accurate model. Reported values from different sources vary markedly due to the technical problems associated to testing small specimens (Anusavice, 2003). Moreover, some data are not easily available, especially when considering commercial restorative materials. Table 2 shows the range of elastic properties used in different recent biomechanical models in the literature.

Material	Young's modulus (GPa)	Poisson's ratio
Dentine	15-18.6	0.30-0.32
Enamel	41-84.1	0.30-0.33
Cementum	13.7-18.7	0.30
Cancellous bone	0.345-1.37	0.22-0.31
Cortical bone	10.7-13.8	0.22-0.31
Periodontal ligament (PDL)	0.05-0.0689 (*)	0.45-0.49
Resin composite	6-20	0.24-0.35
Cement	4.5-95	0.22-0.35
Glass fibre	20-45	0.25-0.30
Stainless steel	207-210	0.30
Gold alloy	93-120	0.33-0.44
Titanium alloy	107	0.34
Gutta-percha	0.00069(*)-0.1	0.30-0.49
Porcelain	68.8-69	0.28
Amalgam	13.72	0.33
Zirconia	205-210	0.30-0.31
Ceramic	62-380	0.25-0.31

(*) With respect to the value of $6.89 \cdot 10^{-2}$ GPa for the elastic modulus of the PDL and the value of $6.9 \cdot 10^{-4}$ GPa for that of gutta-percha, used in many works, a recent paper (Ruse, 2008) has argued that these data are erroneous and have been disseminated by hundreds of work using FEA in recent years as a consequence of a lack of rigour when it comes to citing and checking original papers. Following Ruse, correct values for the PDL are three orders of magnitude lower and for gutta-percha at least two orders of magnitude greater.

Table 2. Range of values for elastic properties for dental materials used in some FE models (Ferrari et al., 2008; Ferrari et al., 2008; Garbin et al., 2010; Gonzalez-Lluch et al., 2009b; Hsu et al., 2009; Maceri et al., 2009; Mezzomo et al., 2011; Okamoto et al., 2008; Schmitter et al., 2010; Suzuki et al., 2008; Toparli, 2003).

In almost all FE models in the literature, the materials of the natural tooth (dentine and enamel) are considered homogeneous, isotropic and elastic. However, in a critical review of the literature Kinney et al. (Kinney et al., 2003) showed that dentine exhibits an hexagonal anisotropy in the Young's modulus with a greater value (36 GPa) in the direction of the tubules than in the orthogonal direction (29 GPa). Later, Ferrari et al. considered this with an orthotropic model for dentine in the FE model (Ferrari et al., 2008). Kinney et al. (Kinney et al., 2003) also reported that dentine, at small deformations, displays a viscoelastic response, i.e. its deformation depends not only on the load that are applied but also on the time since application. Viscoelastic materials, under a constant stress, continue to deform with time, thereby presenting a time-dependence in the mechanical response. However, little is known about how to characterise this viscoelastic behaviour and in the range of frequencies of physiological interest (0.1-10 Hz) Kinney et al. concluded that dentine can be treated as a perfectly elastic solid.

The PDL is a component that surrounds the root of the tooth and connects it to the bone. It contributes to load distribution and damping from the tooth to the bone. Its mechanical behaviour is anisotropic as a consequence of its composition, with collagen fibres that run in different directions (Zhurov et al., 2007). Between the collagen fibres there are tissue fluids that provide a damping behaviour, which enables the ligament to offer a viscous and time-dependent response (Komatsu, 2010). Some works have shown that the elastic response of the PDL is nonlinear (Pini et al., 2002). The elastic modulus is lower for small strains and increases suddenly for strains greater than near 20%-30% in both tension and compression (Pini et al., 2002). However, some models of the ligament in endodontic simulation simplify this characteristic using an isotropic, linear and perfectly elastic material (Garbin et al., 2010; Mezzomo et al., 2011; Okamoto et al., 2008). This simplification avoids the iterative procedure associated to a non-linear model that entails an important increase in computation time. Rees and Jacobsen (Rees & Jacobsen, 1997) analysed the optimal value of the Young's modulus of the PDL (when using an isotropic and linear constitutive model) which provides a good correlation between experimental results and computations of the FE model. They found an optimal value near 50 MPa. This value is used in several works (Davy et al., 1981; Dejak et al., 2005; Garbin et al., 2010) and a similar value of 68.9 MPa in others (Asmussen et al., 2005; Gonzalez-Lluch et al., 2009b; Holmes et al., 1996; Pegoretti et al., 2002). Some recent works consider the non-linear mechanical response of the ligament using a non-linear stress-strain relationship for the finite elements representing the PDL (Okada et al., 2008; Suzuki et al., 2008; Uddanwadiker et al., 2007). Maceri et al. (Maceri et al., 2009) used a set of different spring elements connecting dentine to the bone to represent the non-linear response of the ligament. In an attempt to represent orthodontic movements better, Qian et al. (Qian et al., 2001) proposed the use of special finite elements including fibres that connect two opposing faces of the element. In a further step, some recent works (Ferrari et al., 2008; Sorrentino et al., 2009) considered a more complicated model that takes into account the non-linearity and the time-dependent response of the ligament, using different stress-strain curves for different values of strain rate. Including viscoelasticity in the model is important to be able to simulate the response under dynamical loads correctly, but the fact that most of the models are used in static simulation explains its absence in the majority of the works. The inclusion of the non-linearity and viscoelasticity of the ligament in the constitutive formulation of customised finite elements has been investigated in recent years (Natali et al., 2004; Zhurov et al., 2007), but to the authors' knowledge it has not yet been implemented in a simulation of endodontic restorations. Maceri et al. (Maceri et al., 2010) proposed recently a method to model the PDL and other materials composed of soft collagen tissues, by means of a nano-scale model and a two-step micro–macro homogenisation technique.

Composite posts, such as glass-fibre or carbon-fibre, exhibit anisotropic behaviour due to the presence of reinforcing fibres oriented in the direction of the post axis. An orthotropic model, with different elastic moduli in the direction of the fibres and in the transverse direction, has been used by several authors to represent this material in FE models of the restored tooth (Barjau-Escribano et al., 2006; Garbin et al., 2010; Maceri et al., 2009; Schmitter et al., 2010; Sorrentino et al., 2007).

3.5 Boundary conditions

Boundary conditions in FEA define the known constraints and forces acting on the system. In endodontic simulation, forces represent the loads acting on the tooth and constraints are limitations to the movement of some parts of the model boundary, which are necessary to obtain a unique solution for the displacement of the different points of the model. A clear definition of the boundary conditions that are employed should be included in the description of any FE model, as any changes in these conditions could introduce important differences in the results. Boundary conditions should represent the actual loads and constraints imposed on the real system as well as possible. In this section, we present a review of how these boundary conditions are defined in the biomechanical models described in the literature.

Most of the models presented in the literature limit the analysis to only one tooth. Some works include a portion of the alveolar bone surrounding that tooth. In these cases, restrictions are imposed on the external surfaces of the solid representing the bone (Gonzalez-Lluch et al., 2009b; Hsu et al., 2009). Other works do not include the bone and consider the outer surface of the PDL to be fixed (Ichim et al., 2006; Maceri et al., 2009). Finally, some models do not include the PDL and constrain the system in the external nodes of the root (Lanza et al., 2005; Sorrentino et al., 2007; Zarone et al., 2006). However, this last solution is not recommended because several works have shown that omission of the PDL implies significant changes in stress distribution (Cattaneo et al., 2005; Davy et al., 1981; Qian et al., 2001). Some studies try to reproduce *in vitro* experiments on endodontically restored teeth using FE models. As the rule of thumb in the experimental setup is to substitute the bone by a resin block, the models used in those studies introduce the constraints in the outline of the block of resin, which is attached to the testing machine in the experiment (Schmitter et al., 2010; Soares et al., 2008b).

To reduce the size of the model or in order to have an easier way to show stresses and strains in the sagittal section of the restored tooth, some works include only half of the tooth in the model by eliminating the mesial or distal half (Ferrari et al., 2008; Schmitter et al., 2010). In those cases, a symmetry boundary condition is imposed on the nodes of the cutting plane, thus preventing its movement out of this plane. Although this simplification could be an interesting way to reduce computational time, the simulation of the total tooth is more realistic, as the tooth is not geometrically symmetric.

A considerable number of works in the literature make no explicit reference to the constraints in the paper (Asmussen et al., 2005; Holmes et al., 1996; Nakamura et al., 2006).

The second type of boundary conditions correspond to forces applied to the system. Oral loads can be variable and several simulations can be performed over the model to simulate different oral situations such as occlusion, grinding, mastication of sticky food, impacts on the tooth, bruxism, and so forth. Most FE models of endodontic restorations simulate a maxillary incisor and introduce a load inclined at 45º to the root axis applied to the palatal surface of the tooth in the vestibular direction to represent the direction of oral loads in type I occlusion (Coelho et al., 2009; Hsu et al., 2009; Mezzomo et al., 2011; Okamoto et al., 2008). Other angles close to this value of 45º, ranging from 30º to 60º, have been used in other works (Barjau-Escribano et al., 2006; Ferrari et al., 2008; Garbin et al., 2010; Ichim et al., 2006; Lanza et al., 2005; Zarone et al., 2006). Apart from masticatory loads, other oral conditions have also been simulated, including impacts, using an angle of 90º to the long axis of the tooth, or bruxism, using an angle of 0º (Genovese et al., 2005; Pegoretti et al., 2002). As far as the position of the applied load is concerned, it is typically applied on the palatal surface, below the incisal edge, near the junction of the incisal third and the medial third (Hsu et al., 2009; Li et al., 2006), distal to the cingulum (Ichim et al., 2006). It is important to note that few details are given in the papers about the size of the area over which the force is applied,

although this factor can introduce local changes in stresses and strains near the application zone. Therefore, the authors recommend reporting this information in future works. Two different approaches are used in the selection of the magnitude of the loads applied to the model. Since many of the models are linear, an arbitrary small load of 1 N (Davy et al., 1981) or 10 N (Sorrentino et al., 2007) can be used to analyse the system and to obtain information about the stress distribution, which can be translated to greater force scenarios by a simple scaling operation. This approach is valid to compare stresses among different restorative designs or with respect to the natural tooth. However, other works use a force that represents the expected oral forces in order to obtain stress values that are directly comparable to the stress limits for restorative materials. In these cases, forces of 100 N (Gurbuz et al., 2008; Mezzomo et al., 2011), 200 N (Nakamura et al., 2006; Uddanwadiker et al., 2007) or 300 N (Gonzalez-Lluch et al., 2009b; Hsu et al., 2009) are typical.
For premolars, forces at 0º or 45º were used on the crown (Maceri et al., 2009; Toparli, 2003). Ausiello et al. (Ausiello et al., 2002) distributed an axial load over two different points on the crown using two rigid bars. Okada et al. (Okada et al., 2008) used the data obtained experimentally with a device to measure masticatory forces in vivo to decide the magnitude and direction of the loads in a model of a first premolar. During the experiment a piece of beef jerky was masticated and the resulting load was predominantly directed in the apico-coronal direction (164.3 N) with lower and similar components in the mesial-distal direction and bucco-palatal direction (-28.9 N and -23.9 N respectively). For molars, Imanishi (Imanishi et al., 2003) used three different forces of 225 N to simulate masticatory loads with angles of 0º, 45º and 90º with respect to the radicular axis, and applied it in outer incline of the buccal cusps.
Restored teeth are also subject to thermomechanical loading, because of the transient changes in the temperature of the different components of the restoration when hot or cold foods or liquids are introduced into the mouth. To date very few works included these types of boundary conditions in FE models. Gungor et al. (Gungor et al., 2004) presented a simulation that analysed stresses induced in a first premolar with different all-ceramic crown materials as a consequence of thermal loading.

4. Interpreting FEA results

A correct interpretation of FEA results should be based on the stress and strength of each component in the system. To obtain accurate conclusions from this interpretation, three conditions must be fulfilled: (1) stress values must be reliable, i.e., the FE model should be an adequate representation of the real system, (2) strengths of the different materials present in the model must be known and (3) a suitable failure criterion must be used to compare values of computed stresses, which are bi-axial or tri-axial, with material strengths frequently obtained under conditions of a uniaxial stress state (tension or compression).
The first condition is progressively closer to being fulfilled with the development of more accurate three-dimensional models, with finer meshes and with more components represented in the system. However, as has been commented above, the accuracy of most of the models is still questionable. Many models consider that all materials display linear isotropic behaviour, although this is a simplification for components such as dentine, PDL or fibre posts. A good representation of bonded interfaces is difficult due to their reduced thickness and different ways to model them have been tested, as has been commented in previous sections of this chapter. Moreover, some uncertainty exists about the mechanical

properties of different materials. However, the second and third conditions are not met in most of the previous works, due to the lack of consistent and complete data about the strength of some dental materials. More especially, however, the interpretation of results is hampered by the absence of a post-hoc analysis of failure in each component of the model or the use of the Von Mises equivalent strength as the reference value for this analysis.

To meet the second condition, information about compressive and tensile strengths is needed for each material in the restoration. The compressive strength is usually obtained experimentally by a compressive test using cylindrical specimens. The tensile strength is obtained by applying an axial force over specimens with a cylindrical or rectangular cross-section and is a typical test for metals and other ductile materials. This type of test, however, is rarely used for brittle materials. Technical problems related with gripping and aligning the brittle specimens are often cited as an explanation for not measuring the tensile strengths (Ban & Anusavice, 1990; Xie et al., 2000). Alternatively, the diametral tensile test (DTT) is commonly used to obtain the diametral tensile strength (DTS) (Probster et al., 1997; Xie et al., 2000) because of its simplicity and reproducibility (Coelho Santos et al., 2004). The DTT is performed by compressing a cylindrical specimen with its axis perpendicular to the load direction. Tensile strength can also be obtained indirectly as a flexural strength (FS) with three- or four-point flexural tests (FT) (Probster et al., 1997; Xie et al., 2000). However, DTS and FS are obtained in loading states that are not uniaxial and the results of these tests are not equivalent, as numerous previous works have shown for different dental materials (Ban & Anusavice, 1990; Probster et al., 1997; Xie et al., 2000). Despite this, DTS and FS have been used interchangeably in recent works as a reference to compare with computed maximal stresses in finite element models of dental restorations (De Jager et al., 2006; Imanishi et al., 2003). On analysing the stress state in both tests, FS seems preferable as a reference for the tensile strength of a ductile material because the stress state is uniaxial at the failure point, and this can only be said of a DTT if a plain stress state is assumed, which if far from being the actual situation in real tests.

Finally, the third condition for a correct interpretation of the FE results is related to the selection of the proper failure criterion for comparing actual stress with the admissible stress limits of different materials present in the restoration. A stress criterion or failure theory combines principal stresses at a point in a solid with the compressive strength and tensile strength of the material to obtain a safety factor at this point. Safety factor values lower than unity indicate that the material is prone to mechanical failure at this point and values greater than unity indicate a safe condition at this point. Different criteria such as Von Mises (VM), Rankine (R), Coulomb-Mohr (CM) or Modified Mohr (MM), are usually reported in mechanical engineering manuals for isotropic materials (Shigley & Mischke, 2002). Some of these criteria are better suited for ductile materials, whereas others are more accurate for predicting the failure of brittle materials. In a recent study, Christensen (Christensen, 2006) proposed the use of a novel failure criterion that is valid for ductile and brittle materials.

To date, little research has been devoted to the interpretation of FEM results in endodontic restorations. Most previous work analyzed the results of FE simulations from Von Mises maximal stresses (Asmussen et al., 2005; Boschian Pest et al., 2006; Hsu et al., 2009; Pegoretti et al., 2002; Sorrentino et al., 2007), implicitly assuming the validity of the VM criterion for all the materials used in the restoration. However, it is known that the VM criterion is only valid for ductile materials with equal compressive and tensile strength (De Groot et al., 1987), but materials exhibiting brittle behaviour such as ceramics, cements or resin composites are frequently used in endodontic restorations. Even dentine presents reported values of compressive strength that are significantly greater than its tensile strength (Craig & Powers, 2002). Some authors suggest the use of the Rankine or Maximum Normal Stress criterion to evaluate the failure in dentine, using the maximum principal stress to analyse the results (Ichim et al., 2006; Maceri et al., 2009; Nakamura et al., 2006). Others analyse the results of shear stress at the post-dentine interface, and indicate that this value should be compared to the reported shear bond strengths in order to evaluate the risk of loosing retention (Asmussen et al., 2005; Maceri et al., 2009; Pegoretti et al., 2002). Wakabayashi et al. (Wakabayashi et al., 2008) also underline the importance of shear stresses to anticipate the probability of failure in the adhesive joints of the restoration. Okamoto et al. (Okamoto et al., 2008) used an iterative procedure to account for debonding in the cement layer. Their method involved reducing the elastic modulus of the finite elements of the cement with shear or normal stresses beyond their corresponding strength to a value close to zero and then recalculating the model until they reached a final solution with a reduced adhesion surface in the cement layer.

DeGroot et al. (De Groot et al., 1987) compared three criteria to analyse FEM results in composite resin: Von Mises, a modified Von Mises criterion presented by Williams (Williams, 1973), and the Drüker-Prager criterion. From their results, they concluded that the Drüker-Prager criterion is more suitable for describing the failure of this material. Recently, Christensen (Christensen, 2006) proposed a unified failure criterion for ductile and brittle materials, which is equivalent to the modified Von Mises criterion proposed by Williams with an additional modification for brittle materials. The same author also demonstrated some unrealistic behaviour of the Drüker-Prager and Coulomb-Mohr criteria under certain important stress states.

The interpretation of failure probability in anisotropic materials is even more difficult and few researchers address it. Dejak et al. (Dejak et al., 2007) applied the Tsai-Wu criterion for anisotropic materials to dentine, enamel and resin composites in molars with ceramic inlays.

From the above discussion, it becomes apparent to the authors that more research is needed to clarify the question of how to interpret the results of FE models correctly. In the authors' opinion, DTS should not be used as a reference for the tensile strength in a failure criterion, and FS is a better option. As regards the failure criterion, a critical review is needed about the extensive use of the Von Mises criterion and the use of the novel Christensen criterion should be taken into consideration as an interesting alternative (Pérez-González et al., 2011).

One of the problems related to the interpretation of FE results is the huge quantity of numerical results obtained by the program, namely the displacements and stresses in all the nodes for every dimension considered (two for planar models and three for three-dimensional models). As there may be several thousands of nodes in current FE models, it is not feasible to present the results in a table or a list, at least not for all the nodes. For a good interpretation of the results, one has to know what to look for and where to look for it. It is clear that maximal stresses are the first option to search for points where the system is prone to failure, although the absolute maximum may not be the initial failure point because stress has to be compared to strength in the different components. For a better understanding of the stress distribution in the tooth, some works present the results by giving a list of stresses for selected points or a plot representing the stress values as a function of point location in some trajectory inside the tooth. For example, Davy et al. (Davy et al., 1981) examined the stresses at selected points located near the cervical with different positions in the coronal-apical direction or in the buccal-labial direction. Posterior works (Hsu et al., 2009; Ichim et al., 2006; Pegoretti et al., 2002) also maintain this method, although complemented with fringe plots. Fringe plots are colour representations of some projection of the model using a coloured scale to represent a particular scalar result. Fringe plots allow the distribution of stress or strain in some part of the model to be seen in one picture. They are the usual way to represent the FE results in most works today (Mezzomo et al., 2011; Okamoto et al., 2008; Silva et al., 2009), although they are better suited for qualitative analysis than for quantitative ones. Maceri et al. (Maceri et al., 2009) employed an averaging function to obtain several mean stress values at each section of the tooth in the coronal-apical direction and considered these functions as indicators of the risk of failure. With this procedure, they can easily compare different restoration methods in a plot with simple curves along the axis of the tooth, although the reliability of this customised parameter as risk indicator is questionable.

Statistical tests have been used in some recent works to compare results from different FE models. Hsu et al. used t-tests over sets of points in the areas of interest to compare different restorative solutions (Hsu et al., 2009). The authors have recently presented a similar approach for using ANOVA tests as a method to assess the significance of different restorative designs (Pérez-González et al., 2010). This option for the interpretation of FE results should be investigated in more detail in the future.

5. Validation of biomechanical models

Validation of biomechanical models employed in numerical simulations should be carried out in order to ensure the validity of the results that are obtained. In other biomechanical ambits, model validation is considered to be of major importance (Dalstra et al., 1995; Gupta et al., 2004). Despite this, very few of the works that make use of FEA for endodontic restoration research include such validations of their models. Moreover, little work has been carried out on validation procedures in this particular field.

Generally speaking, two stages should be accomplished in order to validate the FE models. First, some sort of convergence tests have to be performed, where, according to the FEM theory (Zienkiewicz & Taylor, 1989), subsequent refinements of the mesh should make the results to converge. Second, the numerical results have to be compared with those obtained from experimental tests. This comparison can be achieved in different ways that can be found in the scientific literature. Some works only perform the convergence test for validation (Garbin et al., 2010; Sorrentino et al., 2007) and even in theses cases no detailed information about this stage is offered.

One way to accomplish validation of the model is to compare the relation between the load applied and the displacement obtained from both the numerical and the mechanical tests. In some cases (Ausiello et al., 2001), mechanical tests can be undertaken using a testing machine with a constant rate of displacement, which allows two continuous curves to be compared. In other cases, the comparison is achieved only for some discrete values of load and displacements values (Rappelli et al., 2005). In other works the variables that are compared are the load and the strain values (Magne & Tan, 2008; Tajima et al., 2009) using strain gauges, although the process may be found to be more difficult and tedious. In the work by Tajima et al., the utilisation of the strain gauge technique allows simultaneous comparison of the strain values obtained from FE calculation and from experimental tests at four different points, which can be seen as a more rigorous validation. Furthermore, six occlusal points are concurrently loaded during experiments.

An indirect form of experimental validation may be established when the numerical results of stress distribution obtained by the FE model agree with the results obtained by fracture experiments (Barjau-Escribano et al., 2006; Genovese et al., 2005; Huysmans & Van der Varst, 1993). In this sense, model zones with higher stresses should correspond to a higher probability of reaching the failure at these zones during fracture tests when the same external load is applied. Hence, the fracture modes observed during experiments can be compared with the stress distributions patterns obtained by FEA. In other cases, although both FEA and mechanical tests are implemented (Lang et al., 2001; Schmitter et al., 2010; Soares et al., 2008a), the results are considered complementary and no direct validation of the model is pursued.

When FEA is just a part of a more complex numerical model (Maceri et al., 2007), validation for the whole model has to be performed. In this particular case, validation was carried out by comparison with *in vivo* experimental data found in the literature.

6. Conclusions from previous research with biomechanical models

Conclusions from comprehensive three-dimensional models, that capable of simulating the highly irregular shape of real teeth and the real 3D mastication forces acting on them, are summarized in the following section. These models have mainly contributed to the assessment of the effects of the material and the dimensions of the prefabricated posts on the static biomechanical performance of restored teeth, although other parameters have also been analysed.

6.1 About the effect of post material

A conclusion that can be drawn from different works is that stresses distributed differently in natural teeth than in restored teeth and for the case of restored teeth, the distribution is affected in an important way by the material used for the post (Adanir & Belli, 2007; Coelho et al., 2009). The origin of this different distribution lies in the difference in the elastic modulus of the material (near 18 GPa for dentine, 30 GPa for glass fibre posts, and 210 GPa for stainless steel). Stress concentrates where non-homogeneous material distributions are present, just like interfaces. Interfaces of materials with different moduli of elasticity represent the weak link of restorative systems, as the toughness/stiffness mismatch influences the stress distribution (Barjau-Escribano et al., 2006; Sorrentino et al., 2007; Zarone et al., 2006). FE models that consider teeth under flexural-compressive loads concluded that the high elastic modulus of the metal posts caused the stress to be concentrated at the post-dentine junction (Genovese et al., 2005; Hsu et al., 2009; Okamoto et al., 2008; Pegoretti et al., 2002). Consequently, stresses on the root dentine-cortical bone area were weaker than those of the fibre-post group. The results of the FE model of Barjau-Escribano et al. (Barjau-Escribano et al., 2006) made it possible to identify the difference in the elastic moduli between the post and the dentine and core as the origin of stress concentrations at the post-core-cement interface that weakened the restored tooth when stainless steel posts were used. Other works are consistent with this finding (Boschian Pest et al., 2006; Pegoretti et al., 2002). Most of the studies found the highest stresses in restorations with endodontic posts located near the cervical region (Maceri et al., 2009; Mezzomo et al., 2011; Pierrisnard et al., 2002; Sorrentino et al., 2007), especially for fibre posts, whereas peak stresses tend to move apically when more rigid posts are used (Okamoto et al., 2008).

In the FE models from the literature, a monotonic static load was considered, which does not represent the clinical situation, where a dynamic load is characteristic. In a pioneering work, Sancho-Bru et al. (Sancho-Bru et al., 2009) recently proposed a way to use finite element results for a fatigue analysis of dental restorations. From that work, it was concluded that, although restorations using glass fibre posts are able to bear higher static loads, both stainless steel and glass fibre post systems would have a similar life under dynamic loads, although this conclusion should be confirmed using more detailed models in the future.

6.2 About the effect of post diameter

The effect that post diameter has on the stress distribution over the tooth is not the same for all post materials. Several works (Boschian Pest et al., 2006; Nakamura et al., 2006; Rodríguez-Cervantes et al., 2007) have indicated that the effect of diameter is greater for metallic posts than for fibre posts. This could be explained by the fact that the elastic modulus of fibre posts is more comparable with that of dentine, thus producing a more homogenous restoration that is consequently less affected by a change in the diameter of the tooth. In a work by the authors (Rodríguez-Cervantes et al., 2007), we confirmed with *in vitro* tests and FE models that increases in post diameter for metallic posts tend to reduce the failure loads for the restored tooth without the crown. However, in a later work, it was observed that the inclusion of the crown meant that the post diameter did not influence the biomechanical performance of the post systems no matter what post material was used (Gonzalez-Lluch et al., 2009b). For fibre posts, as has been said, the importance of post diameter is lower and the conclusions of different works vary. Boschian-Pest et al. (Boschian Pest et al., 2006), for example, recommended using small diameters to avoid weakening the root, whereas Okamoto et al. (Okamoto et al., 2008) recommended the use of a large diameter, even assuming that changes in stresses are slight.

6.3 About the effect of post length

Two of the earlier FE works (Davy et al., 1981; Holmes et al., 1996) studied metallic post systems by means of two-dimensional finite element models and reached opposing conclusions. The first predicted minor changes in the stress patterns in dentine for the length variations considered; the only effect of post length was a change in the location of the stress concentrations that occurred at the post apex. In contrast, the second predicted higher shear stresses as the length of the metallic post decreased. This controversy seems to continue in two later works that compared the effect of post length for fibre posts. On the one hand, Boschian Pest et al. (Boschian Pest et al., 2006) suggested using a post inserted as deeply as possible, because shorter posts would increase stresses in the luting materials, although minor effects were recognised in the root. Ferrari et al.(Ferrari et al., 2008), on the other hand, concluded that post length does not influence the biomechanics of restored teeth. The same conclusion about the non-significance of the post length was obtained by the authors in a study with *in vitro* tests and with FEA for both metallic and fibre posts (Rodríguez-Cervantes et al., 2007). More recently Hsu et al. (Hsu et al., 2009) concluded that when a metal post is used, the post should be as long as possible, while the biomechanical performance of a glass-fibre post combined with a composite resin core was less sensitive to post length. Others authors, like Chuang et al. (Chuang et al., 2010), using an FE model supported by an *in vitro* study, concluded that post length is more decisive for root fracture resistance in teeth restored with metal posts than in those restored with fibre posts, since long metal posts give rise to stress concentration in the apical portion of the root. They also concluded that post material has a greater effect on the location of peak stress, and on the resulting fracture pattern, than post length does.

6.4 About the effect of other restoration parameters

FE models have corroborated other experimental results and recommendations, e.g., teeth prepared with a ferrule preparation tend to fail in a more favourable mode and exhibit greater mechanical resistance (Ichim et al., 2006) or crowns with small ferrule heights should be resin-bonded instead of using conventional cements (Schmitter et al., 2010). In a recent work, aimed at trying to find the optimal combination of crown material and luting agent, Suzuki et al. (Suzuki et al., 2008) concluded that polymer-based restorative material for the crown and composite cement were preferable to other restorative alternatives. Sorrentino et al. (Sorrentino et al., 2007) also found that changes in the crown and core materials affected stress distribution and that the stress concentrations in post-dentine interface moved apically when more rigid materials were used. The effect of the crown on stress distribution was also studied by the authors and compared with a restoration without the crown (Gonzalez-Lluch et al., 2009b). We confirmed the conclusion reached by Sorrentino et al., although it was also found that the addition of the crown did not affect the final strength of the restoration to any significant extent for either of the post systems considered (stainless steel and glass fibre). As for the effect of the cement properties, Li et al. (Li et al., 2006) found that the elastic modulus of the cement is an important parameter for stress distribution and concluded that cements with an elastic modulus similar to that of dentine should be used in weakened roots.

7. Conclusions and proposals for future work

In the last decades, biomechanical models of endodontic restorations have been developed increasingly using FEA. FE simulations carried out over this time have made it possible to gain a better understanding of how the restored tooth deforms and what stresses it is subject to under simulated loads. These investigations and others to be expected in the future with more comprehensive models could make a valuable contribution to the development of better restorative solutions in this area. With a view to this objective, future works should concentrate on improving current models in order to eliminate remaining weak points. In the authors' opinion, some of these possible future lines of research are:

- Reliable data about the mechanical properties of the different materials used in clinical endodontic restorations are necessary to be able to have good models and interpret results correctly. The stress-strain curves of all these materials, under both tensile and compressive loads, as well as the failure limits, should be clearly established in the literature. Furthermore, common procedures should also be promoted to obtain these data, and communicate and share them among researchers.
- More comprehensive mechanical models will have to be developed in the future. Of course, they should be three-dimensional and represent all the components present in the restoration, but the model should also consider the possible anisotropy of materials, such as bone, dentine or PDL. Additionally, efforts have to be directed towards developing models that represent the nonlinear response of the restored tooth in a suitable manner, because of the nonlinearity of some components, such as the PDL, or due to the appearance of contacts between components.

- Proper validation methods for FE models should be established and shared among researchers. As validation should be based on well-tested and documented experimental results, it is important to increment the quantity and quality of experimental data. To do so, information such as detailed geometrical data about the restored teeth (with different restorative solutions and subject to different loading situations) and experimental measures of strains or displacements should be available to researchers in order to test their numerical models. A common protocol to promote sharing of these data over the Internet would be an important advance.
- Interpretation of FE results is a key point for the future development and reliability of FE models. Of course, the best options to present the results of the models should be investigated, but more research is also needed to establish correct and validated failure criteria for the different materials and especially for bonded interfaces.

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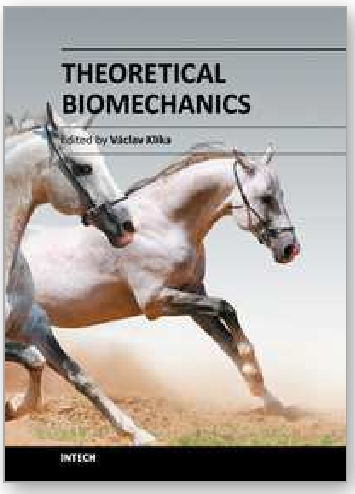

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During last couple of years there has been an increasing recognition that problems arising in biology or related to medicine really need a multidisciplinary approach. For this reason some special branches of both applied theoretical physics and mathematics have recently emerged such as biomechanics, mechanobiology, mathematical biology, biothermodynamics. This first section of the book, General notes on biomechanics and mechanobiology, comprises from theoretical contributions to Biomechanics often providing hypothesis or rationale for a given phenomenon that experiment or clinical study cannot provide. It deals with mechanical properties of living cells and tissues, mechanobiology of fracture healing or evolution of locomotor trends in extinct terrestrial giants. The second section, Biomechanical modelling, is devoted to the rapidly growing field of biomechanical models and modelling approaches to improve our understanding about processes in human body. The last section called Locomotion and joint biomechanics is a collection of works on description and analysis of human locomotion, joint stability and acting forces.