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Evaluation of craniofacial effects during rapid maxillary expansion through combined *in vivo*/*in vitro* and finite element studies

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SUMMARY It is well documented in the literature that a contracted maxilla is commonly associated with nasal obstruction. Midpalatal splitting using the rapid maxillary expansion (RME) technique produces separation of the maxillary halves with consequent widening of the nasal cavity. Although clinicians agree about many of the indications for and outcomes of RME, some disagreements persist in relation to the biomechanical effects induced. The present research was based on the parametric analysis of a finite element model (FEM) of a dry human skull with the RME appliance cemented in place in order to evaluate these effects on the overall craniofacial complex with different suture ossification. The behaviour of the FEM was compared with the findings of a clinical study and to an *in vitro* experiment of the same dry skull. Comparisons refer to the opening pattern and associated displacements of four anatomical points located at the left and right maxilla (MI, UM, EM, CN).

It was found that the maxillolacrymal, the frontomaxillary, the nasomaxillary, the transverse midpalatal sutures, and the suture between the maxilla and pterygoid process of the sphenoid bone did not influence the outcome of RME, while the zygomatico-maxillary suture influenced the response of the craniofacial complex to the expansion forces. Moreover, the sagittal suture at the level of the frontal part of the midpalatal suture plays an important role in the degree and manner of maxillary separation.

Maximum displacements were observed in the area of maxilla below the hard palate, from the central incisors to second premolars, which dissipated at the frontal and parietal bone and nullified at the occipital bone.

Introduction

Rapid maxillary expansion (RME) was first reported by Angell (1860) who used a reciprocal jackscrew attached to premolars as a method of expanding the upper arch. However, it was not until the 1950s that interest in RME was rekindled when Derichsweiler (1953) and Gerlach (1956) reported improvements in nasal respiration in addition to increases in the maxillary apical base due to expansion (Wertz, 1968).

Haas (1961) described a fixed split plate appliance constructed with an acrylic baseplate and a midline expansion screw. The effects of RME on skeletal and dental tissues have been extensively investigated by means of animal studies and clinical research (Haas, 1961, 1965, 1970; Wertz, 1968, 1970; Wertz and Dreskin, 1977; Linder-Aronson and Lindgren, 1979; Timms, 1980; Adkins *et al.*, 1990; Da Silva *et al.*, 1991, 1995). Increases in the transverse dimensions of the upper arch have been attributed to distinct orthopaedic and orthodontic effects (Da Silva *et al.*, 1991). In the first few days of activation of the jackscrew device (RME appliance: Hyrax screw), expansion would be expected to result in compression of the periodontal ligament (PDL) of the posterior teeth included in the device (Bishara and Staley, 1987; Ladner and Muhl, 1995). This in turn would cause bending of the alveolar bone and tipping of these anchor teeth (orthodontic effect). Continued force would cause the suture to gradually open and the maxillae to move away from each other (orthopaedic effect). As the posterior teeth are carried laterally by the separate maxillae, the orthodontic effects produced by RME may be difficult to dissociate from orthopaedic movement of the maxillae.

Forces generated during RME have been investigated by a number of researchers. Isaacson *et al.* (1964) and Zimring and Isaacson (1965) estimated that the force produced by a single turn of a jackscrew in a Haas type appliance was between 3 and 10 lbs, with multiple turns generating loads in excess of 20 lbs. They also reported that the maximum load produced by a jackscrew occurs at the time of activation and begins to dissipate soon after. These heavy forces were considered advantageous to achieve lateral positioning of the maxillae while limiting the amount of tooth movement (Isaacson *et al.*, 1964; Bishara and Staley, 1987). Those authors suggested that the decay following activation is rapid, and that the rate of decay rapidly decreases, while the displacements seem to remain constant. On the other hand, the nearly constant presence of residual forces and the apparent cumulative nature of these forces suggest the possibility that the forces produced by the third and fourth activation are probably greater than the forces present later in treatment. This phenomenon, which is mechanically determined as 'relaxation', plays a key role in the response of the craniofacial complex (Provatidis *et al.*, 2006).

Haas (1961) and Krebs (1964) stated that the shape of the void created by separation was pyramidal, with the base of this pyramid being at the occlusal level and the apex located within the nasal cavity. Wertz (1970) and Timms (1980) reported that at the level of the midpalatal suture, the separation was wider anteriorly at ANS and decreased significantly towards the posterior part of the suture, close to the horizontal part of the palatine bone. Wertz (1970) also found that the maxilla was displaced forward 1.5 mm and downward 1–2 mm. In response, the mandible rotated downward and backward and in all cases an increase in the mandibular plane angle was observed. Moreover, as the suture opened, a diastema (on average) 5 mm (or about half as great as the expansion of the screw) appeared between the crowns of the central incisors.

The anatomy of the midface is complex and the maxillae articulate with 10 other bones of the facial structure and with the anterior and middle cranial base. Starnbach *et al.* (1966) demonstrated that during the early phases of RME, the frontonasal, zygomatico-temporal, and zygomatico-maxillary sutures all showed signs of increased cellular activity. Kudlick (1973) proposed that the pterygoid plates of the sphenoid were the source of the greatest resistance to lateral movement of the maxillae. Although bilaterally placed, they are a single structure without a suture and hence can only bend as a result of expansion. The zygomatic complex also offers some resistance to expansion (Haas, 1961), but these structures are thought to remodel and adjust to their expanded positions (Krebs, 1964).

The effect of RME on the nasal cavity has been reported either as a change in the transverse dimension of the nasal cavity or as an alteration in nasal function as indicated by nasal airway resistance (McDonald, 1995). Wertz (1970) suggested that because of anatomical considerations of the lateral wall of the nose, the effect of RME on the nasal cavity would be limited to the anterior and inferior portion of the nasal cavity, i.e. that part mainly composed of the maxillae. Most investigators (Haas, 1961; Isaacson *et al.*, 1964; Zimring and Isaacson, 1965; Inoue, 1970) however, agree that RME with midpalatal splitting can be accomplished in both adolescents and adults. However, with advancing maturity, the rigidity of the skeletal components limits the extent and the stability of the expansion, and RME in adults may involve fracture of the bony interdigitations. Therefore, the optimal age for expansion is before 13–15 years. The duration of the treatment and the choice of appliance are dependent on the orthodontist.

Although clinicians and researchers agree about many of the indications for and outcomes of RME, the possible mechanism of maxillary expansion is not yet clearly understood. In order to explain the mechanical consequences of RME, various non-invasive computer models have been used to derive conclusions about stresses and displacement distributions on the craniofacial complex: photoelastic models (Chaconas and Caputo, 1982) and three-dimensional (3D) finite element (FE) skull models (Tanne *et al.*, 1989a,b; Işeri *et al.*, 1998; Verrue *et al.*, 2001) have all been investigated. When using the FE method to study a clinical situation, certain assumptions need to be made in order to construct the model. Some assumptions significantly influence the accuracy of the results, including the detailed geometry, material properties and boundary conditions of the finite element model (FEM). Hence, the reliable simulation and evaluation of the mechanical changes within the craniofacial complex of a human skull remains controversial.

The aims of the present study were to systematically investigate RME by means of a FEM. The role of the sutural network of the craniofacial complex and the degree of its ossification on the maxillary segment separation during RME were studied and the results of the finite element analysis (FEA) were compared with the clinical findings of a previous study and an experimental *in vitro* application of the method. Moreover, the way that the maxillary halves move away from each other (orthopaedic effect) as well as the stress–strain field within the PDL and anchor teeth (orthodontic effect) were analysed.

Materials and methods

An adult human dry skull with the maxillary alveolar crest well represented but with the teeth missing, without the mandible and with no evident craniofacial anomaly, was used to construct the FEM. The anatomy of the cranium was obtained using computer tomography (CT).

In order to develop a FEM, it is necessary to transfer the CT images into a 3D computer-aided-design (CAD) and computer-aided engineering (CAE) model. This procedure can be performed faster and more reliably when using dedicated commercial software tools. In the present study, the Mimics/Medcad® (Mimics®: Materialise's Interactive Medical Image Control System, Materialise BV, Leuven, Belgium) was used. This package is capable of reading all types of medical images and transforming them to a stereolithography (STL) format. The STL file was then imported into the CAD/CAE commercial software package of Algor version 12.10 (Algor®, Pittsburgh, Philadelphia, USA). Figure 1 shows the workflow diagram of the generation of the 3D FEM.

The FEM of the derived skull included 75 444 nodes and 1 582 833 elements and a significant number of invalid nodes. Consequently, this automatically obtained model

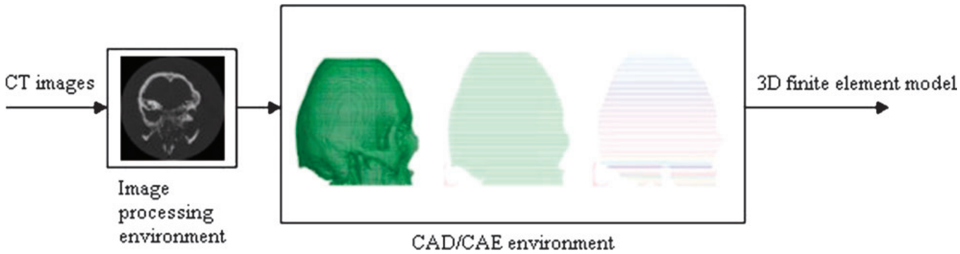


Figure 1 Workflow diagram for the generation of the three-dimensional finite element model.

was not suitable for further investigation. Therefore, from the initial 98 sections, 51 sections were selected automatically by the software in order to construct a new model with the following two criteria (Kotinas, 2007):

1. First, each section was defined by a number of points with a common *z*-vertical co-ordinate. The distance between two succeeding sections was 3 mm except for the nasomaxillary region where the distance was 1.5 mm in order to increase accuracy in the area of application of the expansion force.
2. Second, the true anatomy of the nasomaxillary complex was respected.

Each selected section was divided into FEs interconnected with straight lines that followed the initial anatomy of the skull. For simplicity reasons, the division of each section into FEs was based on the outer geometry of the skull. The construction of the mesh was carried out automatically, using Algor, by converting the lines into splines, and manually where the anatomy included complex and overlapping structures. The various structures were then divided into different groups based on their mechanical properties. The separation of the various bones was at the level of the sutures.

The bone was idealized with shell elements of variable thickness. The acrylic part of the Hyrax jackscrew device was modelled using eight-node brick elements and the metallic part soldered to bands on the anchor teeth using shell and beam elements (Figure 2a). The entire sutural network of the craniofacial complex was modelled with shell elements of variable thickness. Shell elements were also used in order to model the maxillary dentition. The whole model (Figure 3) consisted of 5288 nodes and 5697 elements (Provatidis *et al.*, 2006). All materials were assumed to be linearly elastic, homogenous, and isotropic. For the stainless steel parts, the elastic modulus was chosen as E = 206 840 MPa, for the acrylic device E = 2000 MPa and for the bony structures E = 13 700 MPa (Tanne *et al.*, 1989b; Verrue *et al.*, 2001). Poisson's ratio for all the elements was taken as equal to $\nu = 0.3$. The loading consisted of a given displacement (u) equal to 7.5 mm (each turn of the jackscrew corresponds to 0.25 mm) and was applied on the mesial ends of the symmetric acrylic halves of the jackscrew device in the area of the connecting screw. Figure 2b shows the three couples of nodes at which the given displacement u was imposed. At these nodes, movement was considered only on the x-axis (the two halves of the appliance move away from each other). Therefore, no contact analysis was performed. The model was restrained at the region around the foramen magnum to avoid rigid bodily movement (Tanne *et al.*, 1989b).

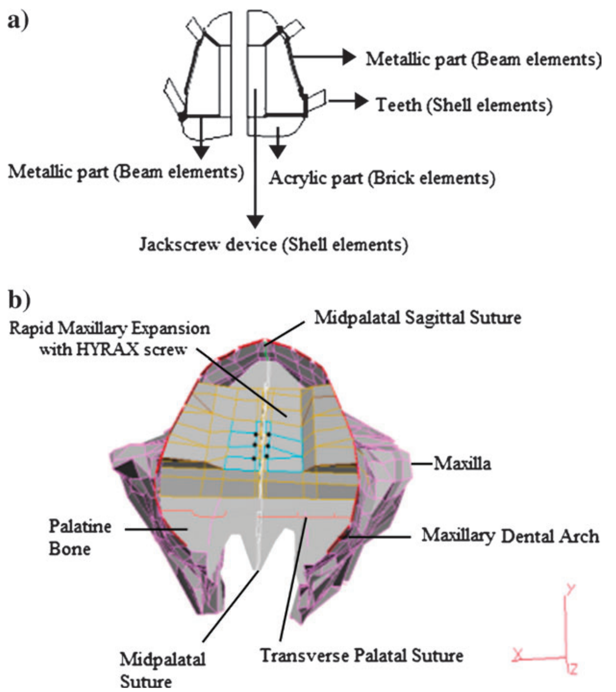


Figure 2 (a) Schematic diagram of the rapid maxillary expansion (RME) appliance used in this study. (b) Finite element model of the maxillae including the RME appliance. The three couples of nodes (black dots) at which the given displacement, u, was imposed are illustrated.

The complexity of bone structures and their interaction with surrounding structures via sutures made it necessary for a number of simplifications to be used (Tanne *et al.*, 1989a,b, 1995; Tanne and Sakuda, 1991; Ruan *et al.*, 1994), including the mechanical properties of the sutures (Verrue

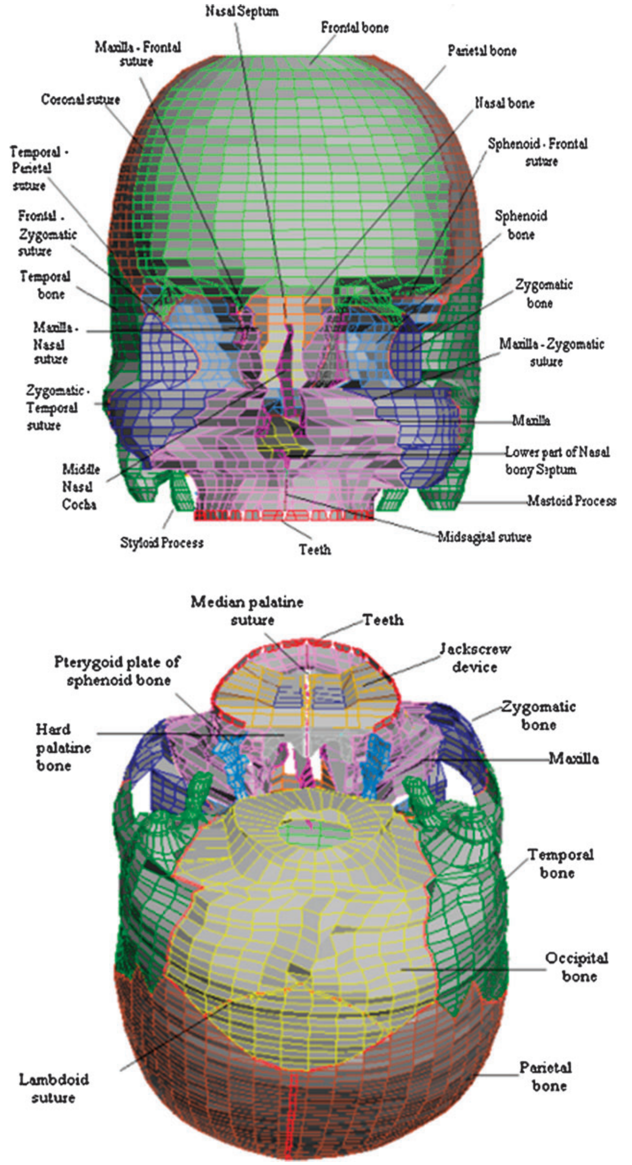


Figure 3 Three-dimensional finite element model of the human skull including sutures, teeth, and the jackscrew device.

et al., 2001). Those authors assigned three discrete situations; unossified, partly ossified, and completely ossified. For every unossified (open) suture, a small modulus of elasticity (E = 1 MPa) (Verrue *et al.*, 2001) was used and, for every ossified (closed) suture, a large modulus of elasticity equal to bone (E = 13 700 MPa) (Tanne *et al.*, 1989b). For partly unossified sutures, a working hypothesis was made, assuming a modulus of elasticity equal to 500 MPa. In all cases, the Poisson's ratio was equal to $\nu = 0.3$.

In order to evaluate the biomechanical effects of RME on the craniofacial/maxillary complex, *in vivo* and *in vitro* measurements were undertaken and compared with the results of the FEA. For the *in vivo* study, 49 patients (average age 13 years 4 months; range 11 years 0 months–15 years 8 months) with various malocclusions who underwent RME as part of their orthodontic treatment were used (McDonald, 1995). They presented with (1) full transverse cusp crossbite; (2) no clinical or radiographic evidence of adenoidal blockage of the nasopharynx; and (3) no previous tonsilar, nasal, or adenoidal surgery.

Kaufman *et al.* (1990) classified the causes of nasal obstruction as mucosal, anatomical or structural abnormalities, and lesions. The mucosal abnormalities include upper respiratory tract infection, allergic rhinitis, rhinitis medicamentosa, and granulomatous disease. Anatomical and structural abnormalities include nasal septal deviation, congenital or traumatic nasal deformity, choanal atresia, the presence of a foreign body, and benign conditions such as allergic polyps, adenoidal hypertrophy, antral-choanal polyp, and juvenile nasopharyngeal angiofibroma. Lateral and postero-anterior (PA) cephalometric radiographs will often reveal some of these conditions. Linder-Aronson (1970), Linder-Aronson and Lindgren (1979) as well as Schulhof (1978) demonstrated the use of lateral cephalometric radiographs to diagnose adenoidal enlargement and blockage of the posterior nasopharyx. However, they recommended that before a diagnosis of nasal obstruction is made, a clinical examination of the nasal cavity should also be undertaken.

In addition to these criteria, for the purposes of this study, all patients presented with complete medical and dental records including good PA cephalograms with the expansion device *in situ* at the end of active expansion. The duration of the active stage of expansion was 10 days. The screw was turned three times daily with each turn producing 0.25 mm of expansion between the two halves of the device. When the required expansion was achieved, the screw was locked in position and was used as a retainer for 3 months.

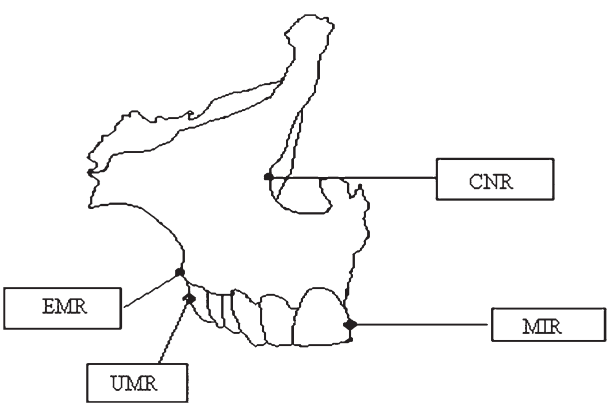
The original lateral cephalometric radiographs were taken with the aid of a cephalostat and a Morito Pan X E2 dental pantomogram. Trimax 3M blue-based fast radiographic film was used in a cassette with a rare earth screen. Exposures were made at 80 kV for 8 seconds. PA radiographs were taken using the same equipment with the patients also in the natural head position. The film was exposed for 1.3 seconds at 80 kV.

Both sets of radiographs, before and after RME, were examined to ensure a clear image of a wide range of skeletal, dental, and nasal structures. A computer-based system was used to digitize and to measure skeletal, dental, and nasal measurements. The software consisted of a commercially available cephalometric analysis program, Viewbox v2.60.02 (Manual version; dHAL©, Kifisia, Greece). This program has an extensive library of cephalometric landmarks and measurements together with the facility to create additional operator-generated landmarks and cephalometric measurements.

Fifty-nine skeletal, dental, and nasal landmarks were digitized and from these landmarks 25 measurements were identified and analysed using a customized analysis within the Viewbox software program. All measurements tested conformed to normal distribution. Statistical analysis of the results showed that only four linear distances on the PA cephalograms presented statistically significant differences ($P < 0.005$). The anatomical definition of these four points was also easily reproducible on the FEM. Therefore, the displacement pattern of these four linear measurements (MIR–MIL, UMR–MIL, EMR–EML, CNR–CNL) (Figure 4) were calculated during the FEA in order to assess the qualitative and quantitative behaviour of the model.

A 99.5 per cent confidence interval for the true difference between the population means (McDonald, 1995) was then calculated (Table 1). If the results of the FEA were found within the limits of the confidence interval, it is possible to define all these parameters that influence the clinical result of RME.

For the *in vitro* experiment, the same human dry skull was used. It was secured to the table surface with a custom-made metal jig. The RME appliance was cemented in place on the first permanent maxillary premolars and molars. These teeth were artificially constructed in the dental laboratory from heat-cured acrylic after taking impressions of the empty alveolus in order to achieve a close fit of the root in the alveolus. Spherical landmarks were used in order to obtain more accurate measurements. The Hyrax screw of the RME appliance was opened in small increments causing a corresponding displacement of the two parts of the jackscrew. It was possible to perform only seven turns, each of 0.25 mm, i.e. a total 1.75 mm of displacement. At this point, it was not possible to continue due to a lack of elasticity in the dry skull (Provatidis *et al.*, 2003).



MIR(L) : The most mesial point on the upper right (left) incisor

EMR(L) : Ecto-maxillare right (left)

CNR(L) : The right (left) cavum nasi

UMR(L) : The most lateral point on the upper right (left) first molar

Figure 4 Definitions of the measuring points used.

Measurements were obtained using a digital micrometre (MDC-75M; Mitutoyo, Kawasaki, Japan, 0.001 mm), an inside micrometre (Mitutoyo, 0.01 mm) and a digital Vernier gauge (0.01 mm). Two video cameras were used to record the whole experiment.

Results

Initially, an attempt was made to study the effect of the degree of ossification of the craniofacial sutures on the behaviour of the FEM of the dry skull. It was considered that the modulus of elasticity for bones was 13 700 MPa and the modulus of elasticity for all the sutures was 1 MPa (model 1). The results are presented in Table 2. Next, it was considered that the modulus of elasticity for bones was 13 700 MPa and the modulus of elasticity for all the sutures was 13 700 MPa (model 2). The results are presented in Table 2. It was observed that the two maxillary halves did not open in a pyramidal shape.

According to Haas (1965), the transseptal fibres that connect the roots of the central incisors have a small modulus of elasticity. Thus, the behaviour of the dental points MIR–MIL may also depend in part on these fibres. Thus, the sagittal suture at the level of the frontal part of the midpalatal suture, which lies between the central incisors from the alveolar crest to ANS, was subdivided into an individual structure (Figure 5). Based on model 2, partly unossified material properties (500 MPa) were assumed for this structure leading to model 3. In addition, the circummaxillary sutures of this model were assumed to be completely unossified (1 MPa). The findings for model 3 are presented in Table 3. It was noted that the two maxillary halves started to open in a pyramidal shape.

Next, a parametric analysis for assessing the role of the various circummaxillary sutures and the other structures of the craniofacial complex was undertaken. Given that there is a different modulus of elasticity (13 700 MPa) at the maxillo-lacrymal (model 4), the frontomaxillary (model 5), the naso-maxillary (model 6) sutures, the suture between the maxilla and the pterygoid process of the sphenoid (model 7), and the transverse palatal suture (model 8), the results remained almost the same. In contrast, a modulus of elasticity of 13 700 MPa at the zygomatico-maxillary suture (model 9) resulted in the changes shown in Table 3.

Clinical studies have suggested that all circumaxillary sutures are totally ossified with the exception of the midpalatal sutures (sagittal and transverse), which are partly or completely unossified in order to apply the RME technique. Consequently, an additional FEM (model 10) was analysed assuming that the sagittal and transverse midpalatal sutures were completely unossified. In other words, it was hypothesized that this model would better

Table 1 The confidence interval for the mean population difference (in mm).

Measuring points	Difference of openings between the mean values before and after treatment (in mm) $\Delta X = \bar{x}_{after} - \bar{x}_{before}$	Confidence interval for the mean population difference ($P = 99.5\%$) in mm
CNR–CNL	1.66	$0.3 < \mu_2 - \mu_1 < 3.01$
EMR–EML	2.47	$0.43 < \mu_2 - \mu_1 < 4.5$
UMR–UML	8.89	$6.77 < \mu_2 - \mu_1 < 11$
MIR–MIL	1.08	$0.49 < \mu_2 - \mu_1 < 1.66$

Table 2 Results of the finite element analysis results when all craniofacial sutures were considered as unossified and ossified.

Measuring points	Displacements derived from finite element analysis (mm)	Difference of openings between the mean values before and after treatment (mm)	Confidence interval for the mean difference of population ($P = 99.5\%$) (mm)
	Sutures: $E = 1$ MPa	$\Delta X = \bar{x}_{after} - \bar{x}_{before}$	
CNR–CNL	5.1	1.66	$0.3 < \mu_2 - \mu_1 < 3.01$
EMR–EML	5.53	2.47	$0.43 < \mu_2 - \mu_1 < 4.5$
UMR–UML	6.1	8.89	$6.77 < \mu_2 - \mu_1 < 11$
MIR–MIL	7.7	1.08	$0.49 < \mu_2 - \mu_1 < 1.66$
	Sutures: $E = 13\,700$ MPa		
CNR–CNL	0.4	1.66	$0.3 < \mu_2 - \mu_1 < 3.01$
EMR–EML	2.1	2.47	$0.43 < \mu_2 - \mu_1 < 4.5$
UMR–UML	2.6	8.89	$6.77 < \mu_2 - \mu_1 < 11$
MIR–MIL	0.08	1.08	$0.49 < \mu_2 - \mu_1 < 1.66$

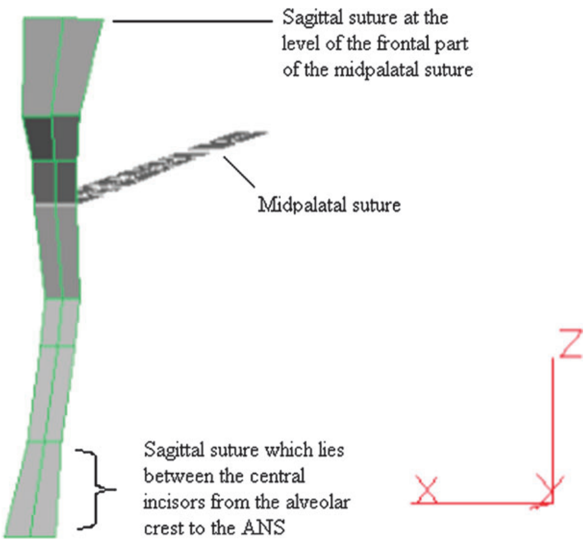


Figure 5 The sagittal/midpalatal suture.

simulate the clinical situation. The results are shown in Table 3 and the displacement distribution on the overall craniofacial complex in Figure 6. It was found that the qualitative behaviour of the model corresponded to the clinical findings. The maxillae expanded in a pyramidal manner with the base being at the level of the dentition and the apex superiorly in the nasal cavity and with the maximum opening at the level of the incisor teeth and decreasing posteriorly again in a pyramidal manner.

As the main aim of the present study was to decide which FEM was closer to the clinical situation and accepting that the qualitative behaviour of a dry skull does not simulate with a high degree of accuracy the clinical situation, it is possible to indicate the way that the two maxillary halves separate during expansion and how the effect of the force influences the other craniofacial structures. Therefore, it was considered necessary to apply RME on a dry skull in order to acquire more information on the distribution of the displacements, the order of suture separation, and the linearity of the method. At this point, it is important to mention that in FEA the expansion was assumed to be linear and the tissue rebound phenomenon was not taken into consideration in the calculations. Consequently, the behaviour of the model and the dry skull must be very similar.

After only seven turns of the Hyrax screw, each of 0.25 mm, the resistance of the surrounding structures of the nasomaxillary complex to further expansion of the two maxillary halves was extremely high and it was difficult to continue turning the screw. This was probably due to lack of vitality and consequent lack of elasticity of the dry skull. Thus, an extrapolation of the *in vitro* measurements was performed to predict the openings for a higher number of turns. This was carried out using linear interpolation. The results for each critical point are presented in Figure 7.

Table 3 Finite element analysis results when the transseptal fibres between the maxillary central incisors are considered with modulus of elasticity $E = 500$ MPa, the maxillozygomatic suture is considered ossified, and the sagittal–midpalatal sutures are considered as totally unossified.

Measuring points	Displacements derived from finite element analysis (in mm)	Difference of openings between the mean values before and after treatment (in mm)	Confidence interval for the mean population difference ($P = 99.5\%$) (in mm)
	Transseptal sutures: $E = 500$ MPa	$\Delta X = \bar{x}_{after} - \bar{x}_{before}$	
CNR–CNL	2.61	1.66	$0.3 < \mu_2 - \mu_1 < 3.01$
EMR–EML	4.62	2.47	$0.43 < \mu_2 - \mu_1 < 4.5$
UMR–UML	5.15	8.89	$6.77 < \mu_2 - \mu_1 < 11$
MIR–MIL	1.39	1.08	$0.49 < \mu_2 - \mu_1 < 1.66$
	Maxillozygomatic suture: $E = 13\,700$ MPa		
CNR–CNL	2.75	1.66	$0.3 < \mu_2 - \mu_1 < 3.01$
EMR–EML	4.28	2.47	$0.43 < \mu_2 - \mu_1 < 4.5$
UMR–UML	4.76	8.89	$6.77 < \mu_2 - \mu_1 < 11$
MIR–MIL	1.36	1.08	$0.49 < \mu_2 - \mu_1 < 1.66$
	Sagittal–midpalatal sutures: $E = 1$ MPa		
CNR–CNL	1.28	1.66	$0.3 < \mu_2 - \mu_1 < 3.01$
EMR–EML	3.54	2.47	$0.43 < \mu_2 - \mu_1 < 4.5$
UMR–UML	4.25	8.89	$6.77 < \mu_2 - \mu_1 < 11$
MIR–MIL	2.09	1.08	$0.49 < \mu_2 - \mu_1 < 1.66$

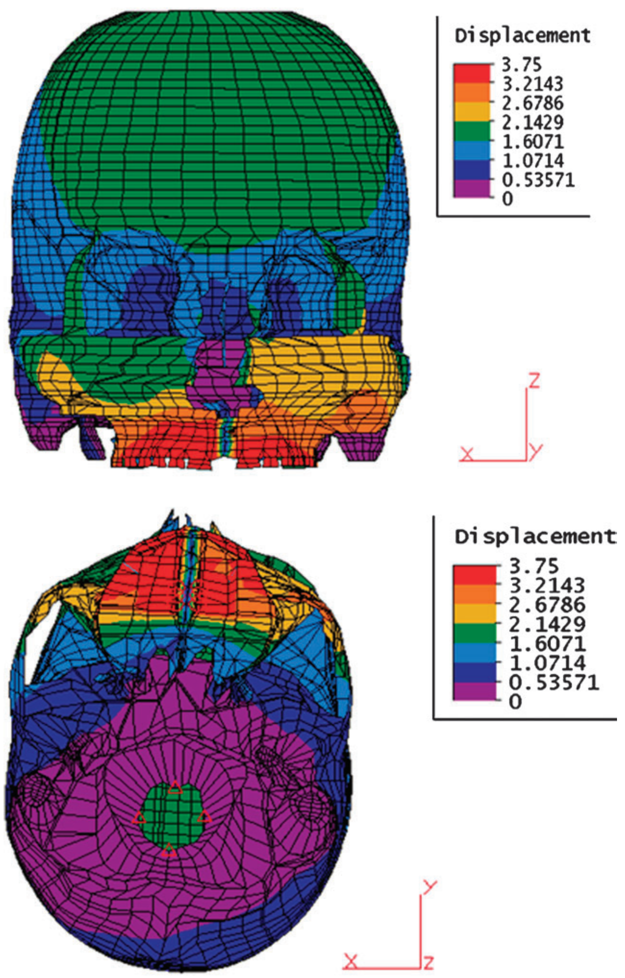


Figure 6 Displacement distribution on the craniofacial complex for model 10 (sagittal and transverse midpalatal sutures completely unossified).

The *in vitro* experiment showed that the maxillary halves separate in a pyramidal manner with the base being at the incisor area and the apex being in the posterior region of the maxillae. This is also in accordance with the clinical findings. The results show that the skeletal points (i.e. CNR–CNL and EMR–EML) were within the confidence interval for the mean population difference. The displacements of the other two dental points were outside the confidence interval which was due to a lack of bone relaxation which does not occur at the dry skull and a lack of simulation of tooth movement within the alveolar crest. A comparison between the *in vivo*, *in vitro*, and the FEM is shown in Table 4.

Discussion

This study was based on the parametric analysis of a 3D FEM and an *in vitro* experiment of the same dry human skull in order to investigate the biomechanical effects on the craniofacial complex during RME. Moreover, the findings were compared with the clinical outcomes of 49 patients (McDonald, 1995).

The FEM is a standard tool used in engineering to precisely assess local stress/strain/displacement distribution in geometrically complex structures. It provides the possibility to simulate orthodontic force systems used clinically and to study biomechanical variables induced in living structures by various external forces in 3D space. The value of FEA entirely depends on element density, accuracy of geometry, material properties, and loading conditions. Compared with other engineering applications, geometric data for most human parts are not available. Furthermore, biological material properties must be converted into numerical material properties suitable for the FE program code. For

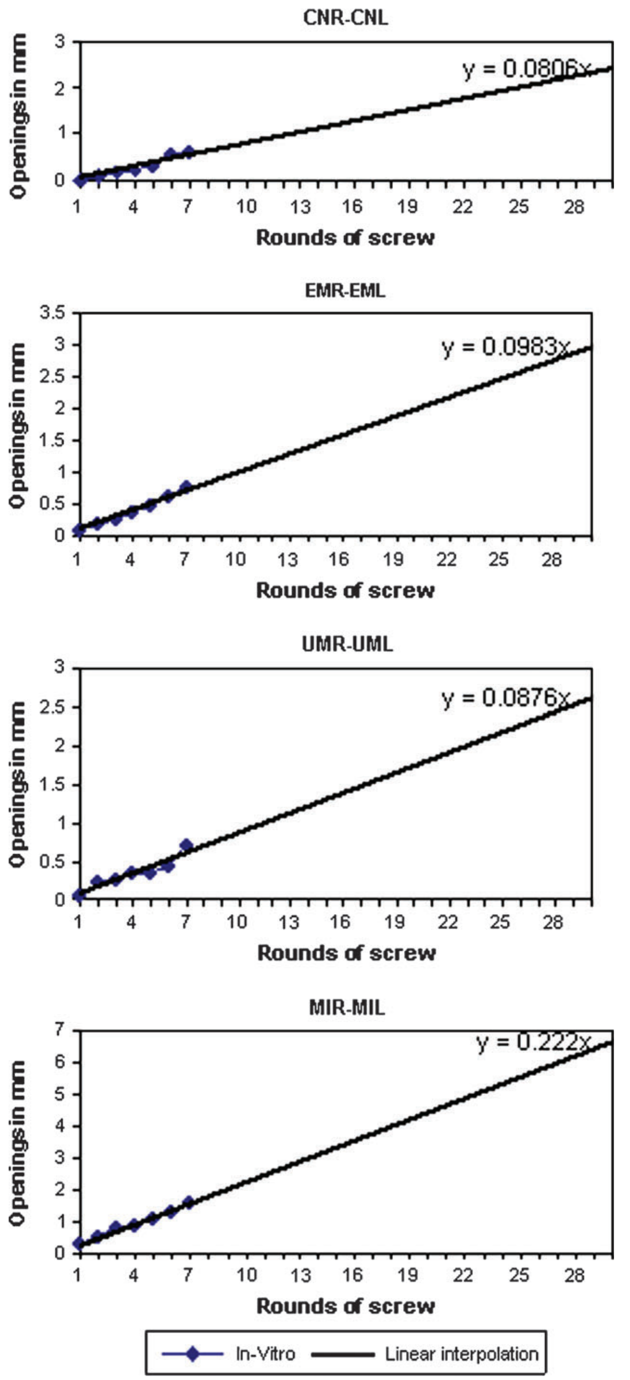


Figure 7 Displacements of all measuring points obtained in the *in vitro* experiment.

the construction of the FEM of the craniofacial complex, a human dry skull was used. The geometry of the cranium was obtained using the CT technique.

CT represents a well-established method of medical imaging with high-quality sagittal sections of the inner structure of the body. It has a number of advantages over previous reconstruction methods, particularly the ability to automatically locate object boundary outlines and produce shaped 3D images of reconstructed data (Moaddab *et al.*, 1985; Hart *et al.*, 1992; Sprawls, 1992; Mehta *et al.*, 1997). Qualitative comparisons between a technique that involves embedding a mandible in plastic resin and cutting serial sections and a technique that involves obtaining anatomy using the CT scans showed that only minor positional errors (of the order of 1 mm) were introduced by the CT scans (Hart *et al.*, 1992).

Commercially available software was used, firstly, to reconstruct the craniofacial complex into a 3D model, and secondly, to divide the model into FEs and construct the mesh. The division of each section into FEs was based on the outer geometry of the skull. The construction of the mesh was done automatically by the software where the geometry of the model was simple, and by converting the lines into splines, and, manually where the anatomy was more complex. Shell elements were used for both the naso-maxillary complex and the other cranial structures, as well as for the teeth. By using a shell mesh, additional elements were available for the biomechanically more important representation of the maxillary dentition and the RME appliance, as well as the craniofacial connecting sutures (Remmler *et al.*, 1998). The maxillary dentoalveolar arch and the model of the RME appliance were added artificially with respect to the proportional relationship of the teeth to the whole skull. In addition, with shell elements, the mathematical calculations were easier when forces were applied perpendicular to their surface. The model consisted of 5288 nodes, 5697 elements, and 3906 degrees of freedom. The objectives for this clinical CT/FEA differed from previous studies in that the predictions sought the overall deformation of a heterogenous nasomaxillary and craniofacial skeleton subjected to force loading from the Hyrax screw of the RME appliance. Moreover, in the present study, displacements were calculated rather than strain and stress.

The human neurocranium remains a geometrically complex structure that features internal voids and branching structures and an extensive network of sutures that interconnect the bones of the craniofacial skeleton. It consists of an external cortical layer and internal cancellous bone which blend one into another rather than demarcating sharply and vary greatly both in relative thickness and density (Melvin *et al.*, 1969; McElhaney *et al.*, 1970; Hubbard, 1971). It also contains various soft tissues with different material properties. Therefore, in order to study all the biological phenomena that take place, a number of assumptions were made in the construction and the analysis of the FEM developed in this study that may, in theory, lead to less precision in the results:

1. In accordance with the assumptions of earlier craniofacial FE studies (Tanne *et al.*, 1989a,b, 1995; Tanne and Sakuda, 1991; Ruan *et al.*, 1994), linearly elastic and isotropic behaviour for the full thickness of bone with a modulus of elasticity equal to 13700 MPa was assumed.

Table 4 A comparison between the *in vivo*, *in vitro*, and finite element analysis results.

Measuring points	Difference of openings between the mean values before and after treatment (mm)	Confidence interval for the mean difference of population ($P = 99.5\%$) (mm)	Openings after extrapolation of the measurements obtained (mm)	Displacements derived from finite element analysis (mm)
	$\Delta X = \bar{x}_{\text{after}} - \bar{x}_{\text{before}}$		*In vitro* study	Sagittal–midpalatal sutures: $E = 1$ MPa
CNR–CNL	1.66	$0.3 < \mu_2 - \mu_1 < 3.01$	1.28	
EMR–EML	2.47	$0.43 < \mu_2 - \mu_1 < 4.5$	2.949	3.54
UMR–UML	8.89	$6.77 < \mu_2 - \mu_1 < 11$	2.628	4.25
MIR–MIL	1.08	$0.49 < \mu_2 - \mu_1 < 1.66$	6.66	2.09

2. The maxilla was considered as a rigid body and consequently any bone bending under force application was not considered.
3. The entire sutural network behaves in an elastic manner, but different combinations of material properties create individual participation in the behaviour of the maxillae.
4. The skull was packed at the cranial base along the foramen magnum (i.e. all degrees of freedom were constrained) and the total mechanical load applied resulted from the expansion caused by the displacement of the midline screw of the RME appliance and was equal to 7.5 mm.

Separation of the maxillary halves was considered to be linear, the mechanical load of each turn of the screw being added algebraically to the pre-existing load. In the clinical situation between each turn of the screw, there is a relaxation period with consequent tissue rebound (Zimring and Isaacson, 1965), which was not considered in the initial pilot testing of the present study. Only displacements were calculated and not strains and stresses because with the boundary condition of the 7.5 mm opening of the Hyrax screw, the von Misses stresses developing in the craniofacial skeleton have been found to reach values as high as 550 MPa (Işeri *et al.*, 1998) which are considerably higher than the yield strength (130 MPa) of the bone (Kayabasi *et al.*, 2006).

Perceived shortcomings of the model included a lack of sophistication in modelling the anatomical structures and maxillary dentition; a lack of detailed knowledge regarding the material properties of the cancellous and cortical bone, the sutures, and the structural parts of the RME appliance; and the difficulty of knowing how to model the boundary conditions at the sutures (Middleton *et al.*, 1996). Therefore, the following discussion is aimed at providing answers to the questions regarding the qualitative behaviour of the maxilla as a result of application of simulated orthodontic–orthopaedic loads; to quantify the resultant displacements and stresses within the maxillary sutures, and to discuss their interaction with biological phenomena such as suture separation, bone remodelling, and tooth movement.

The anatomy of the midface is complex and the maxillae articulate with 10 different bones within the facial structure and the anterior and middle cranial base. This is responsible for the pyramidal shape of the expansion both anteroposteriorly and vertically (Wertz, 1970; Da Silva *et al.*, 1995). Starnbach *et al.* (1966) demonstrated that during the early phases of RME, the frontonasal, zygomatico-temporal, and zygomatico-maxillary sutures all showed signs of increased cellular activity. Haas (1961), as well as Bishara and Staley (1987), suggested that the main resistance to midpalatal suture opening is from the sphenoid and zygomatic bones. Kudlick (1973) maintained that the pterygoid plates of the sphenoid limited the ability of the palatine bones to separate at the median palatine suture. Although bilaterally placed, they are a single structure and do not have a suture. As a consequence, the pterygoid plates can only bend under the pressure of the expansion forces.

The results obtained with the present FEM, even though boundary conditions were at the upper and lower limits of the clinical situation for easier identification of the changes produced, were qualitatively similar with the above studies. The maxillae opened in a pyramidal shape with the base being at the incisor area and the apex being in the posterior region of the maxillae. Maximum displacements were measured at the maxillae below the hard palate from the central incisors to the second premolars and, thereafter, dissipated at the deeper anatomical structures to become zero at the occipital bone. Minimum displacements were calculated at the pterygoid plates close to the cranial base (Jafari *et al.*, 2003).

Very little attention has been devoted to experimental validation of a FEM. The validity of the model can be addressed in several ways, to include in-depth investigation of the structure by parametric analysis (Huiskes and Chao, 1983). The influences of the parameters on its functional performance are often difficult to define quantitatively. Modelling aspects can be evaluated by changing the descriptions of loading, material properties, and boundary conditions and repeating the calculations.

Earlier studies showed that the varying degree of ossification of the circummaxillary sutures of the craniofacial complex (Melsen and Melsen, 1982) resulted in varying degrees of displacement of the various structures (Isaacson *et al.*, 1964; Gardner and Kronman, 1971; Timms, 1980; Vardimon *et al.*, 1998). Assuming that the mechanical properties of bones are constant (E = 13 700 MPa), the degree of ossification can be simulated by conferring different mechanical properties to the sutures which unite the bones. Since the exact mechanical properties of sutures at each stage of human development are not well defined in the literature, this research included a parametric analysis of the circummaxillary sutures at two extreme stages: firstly, when all sutures of the craniofacial complex were considered unossified (E = 1 MPa) and secondly when totally ossified (i.e. advanced maturity; E = 13 700 MPa). The third one was assumed to be partly unossified leading to the construction of a working hypothesis that was E = 500 MPa.

In the first case of unossified sutures (model 1), the results showed an extreme separation of the maxillary halves, beyond the clinical findings, with all measurement points found outside the confidence interval of population means. For ossified sutures (model 2), the displacement distributions decreased significantly with two of the measuring distances found within the confidence interval (i.e. CNR–CNL and EMR–EML). When an attempt was made to simulate the effect of the transeptal fibres between the maxillary central incisors (model 3) as partly unossified, the considerable difference that was initially noted between points MIR–MIL changed significantly, demonstrating the major influence of these fibres on the creation of a midline diastema during RME.

However, since the various sutures are at different stages of development, alternative models were used to test the qualitative and quantitative behaviour of each suture separately under the loading conditions of the expansion screw of the RME appliance. The results of this parametric analysis showed that the maxillo-lacrymal, the fronto-maxillary, the transverse palatal suture, and the suture between the maxilla and the pterygoid process of the sphenoid bone do not influence the FEA results. On the other hand, it was demonstrated that the zygomatico-maxillary suture influences the displacements of the measuring points used in this study. In conclusion, model 10, which was assumed to have all craniofacial sutures ossified, with the exception of the palatal (sagittal and transverse) sutures, proved to simulate the clinical situation in a more reliable way in order for maxillary expansion to occur.

Computer studies on models as well as experiments with dry skulls cannot, long-term, replicate the reactions of living tissues, only predict it. Short-term, however, there is only mechanical force involved. Consequently, since the radiographic records of the clinical subjects were taken at the end of the active phase of expansion, before complex biomechanical interactions between the various living components of the craniofacial complex result in significant remodelling, this comparison is feasible.

Comparing the results (Table 4), it was shown that the displacements of the skeletal points both of the *in vitro* and FE study were inside the confidence interval for the mean difference ($P < 0.005$). On the other hand, the displacements of the dental points were outside the confidence interval due to tooth tipping within the PDL and alveolar bone bending, both of which do not occur in the dry skull and are not considered in the FEM. Alveolar bending occurs in living tissues due to their elasticity, a property that is the result of a complex interaction between various biological structures that cannot yet be accurately transferred to a computer model. Moreover, contact and friction effects within the jackscrew device were not considered in this model. Therefore, a smaller opening for the UMR–UML (dental points) and a larger opening for the EMR–EML (skeletal points) was observed in comparison with the clinical findings (Figure 8). Even when an attempt was made to model the PDL according to present knowledge and the central processing unit possibilities, a considerable difference between the *in vivo* measurements still existed because of the small thickness of the PDL and insufficient scientific knowledge about its exact material properties. In the case of the MIR–MIL measurements, an additional explanation may be the overlapping of the maxillary central incisors, a common finding in clinical cases before treatment due to a narrower dental arch. This overlapping decreases the absolute value of the displacement between these dental points on the PA radiographs.

The conclusions that can be drawn from this study are only useful in understanding the qualitative behaviour of the craniofacial complex during RME. However, qualitative

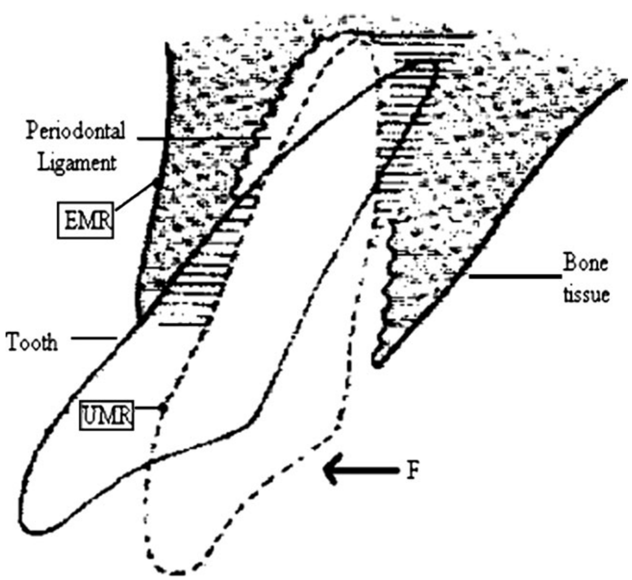


Figure 8 Schematic presentation of the direction of displacements for points EM and UM under the influence of the expansion forces of the jackscrew device. The dashed line corresponds to the displacements simulated by the present finite element model where the teeth are assumed with no movement within the periodontal ligament.

research has its own scientific value. It can be used to generate hypotheses in the development of patient-based measures and in explaining research findings and informing future investigations.

Conclusions

It appears feasible to simulate orthopaedic changes in the nasomaxillary complex by means of FEM. Nonetheless, as emphasized in recent reviews of cranial and maxillofacial FE studies (Voo *et al.*, 1996; Korioth and Versluis, 1997), expectations of high accuracy in all regions of biologic FEMs are rarely achievable. Therefore, the specific objectives of the study must be matched to available computational resources and the mesh design and boundary conditions optimized to shift the accuracy of the model towards regions of interest.

1. The pyramidal shape of expansion is a result of the different degrees of resistance that the midpalatal suture of the maxilla encounters along its length. An important role is the frontal part of the midpalatal suture, especially at the level of the transseptal fibres.
2. FEA of models that consider the midpalatal suture as unossified and the *in vitro* experiment of the dry human skull both suggest that the maxillary halves in reaction to the expansion forces of the jackscrew device of RME appliance separate in a pyramidal manner with the base being at the incisor area and the apex being in the posterior region of the maxillae. In the vertical dimension, maximum opening occurs at the level of the dentition and decreases in an upward direction.
3. The maxillolacrymal, frontomaxillary, nasomaxillary, the transverse palatal sutures, and the suture between the maxilla and the pterygoid process of the sphenoid bone do not influence the outcome of RME. On the contrary, the zygomatico-maxillary sutures at the level of the zygomatic arch influence the response of the craniofacial complex to the expansion forces.
4. The degree of sutural ossification plays a significant role in determining the displacement distribution over the craniofacial skeleton. The sutures that separate the maxillary halves from each other must be unossified in order for maxillary expansion to occur.
5. The results show that the maximum displacements are observed in the area of the maxillae below the hard palate and from the central incisors to the second premolars. Viewed frontally, the displacements are dissipated at the frontal and parietal bone and become zero at the occipital bone.

The most significant positive contribution of the FEM is the ability to predict events at sites at which measurements are impossible in living humans. Currently, the approach is limited only by the computational power available to the individual researcher, the availability and reliability of suitable values for the many variables involved, and, in this present case, conditions of static loading. In future studies, larger FE meshes and more measuring points for detailed comparison with clinical findings would be advantageous.

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