JOURNAL OF MATERIALS SCIENCE: MATERIALS IN MEDICINE **16** (2005) 1191–1197

An experimental and theoretical composite model of the human mandible

R. DE SANTIS[1,*], F. MOLLICA[2], R. ESPOSITO[1], L. AMBROSIO[1], L. NICOLAIS[1]
[1] *Institute of Composite and Biomedical Materials-National Research Council, Piazzale Tecchio 80, Napoli 80125, Italy*
E-mail: rosantis@unina.it
[2] *Department of Materials Engineering, University of Ferrara, Via Saragat 1, Ferrara 44100, Italy*

The purpose is to design and manufacture a composite mandible replicate suitable for testing the influence of prosthetic materials on the stress distribution of bone. Composite mandibles made of a poly(methylmethacrylate) core and a glass reinforced outer shell are manufactured and characterised through mechanical tests assisted by the finite element analysis. The mandible replicate has been conveniently equipped with strain gauges, moreover a video extensometer has also been used in order to measure the arch width change during loading.

A close agreement is found between the experimental data and the theoretical predictions. By laterally loading the mandibles the maximum values of stress and strain take place in the premolar-incisal region.

By varying technological parameters such as the fiber volume fraction and orientation, it is easy to replicate the behaviour of mandibles having different stiffnesses. The results obtained by laterally loading the composite mandibles through the condyles or through the gonion regions are consistent with literature data relative to the arch width decrease of natural jaws during opening and closing. This novel synthetic system coupled with the Finite Element model constitutes an experimental-theoretical model suitable to investigate the biomechanical effects of oral rehabilitations on mandibular bone.

1. Introduction

Osseointegrated titanium implants are being used increasingly in dental surgery in order to restore the oral function of edentulous patients. Since integrated implants have a load bearing and a stress transfer function it is necessary to understand the biomechanical behaviour of the mandible rehabilitated with implants. Bone quality is one of the key parameters influencing the success of oral rehabilitation. Particularly, the long term stability of implanted prostheses is strongly related to the stress transfer between implants and bone, in fact both stress concentration and stress shielding lead to catastrophic effects on bone's health (i.e. bone necrosis and resorption) and therefore on the bone-implant stability [1–4]. Therefore, implant loading, stress transfer to bone and the bone-implant interface stability are topics of great interest in current literature. Unfortunately, quantitative data related to these issues are still missing. Nevertheless, a proper knowledge of the mandible biomechanics is essential to improve orthodontics and temporo-mandibular joint treatments designed to solve the specific pathology and disorder.

Explanted mandibles are used in order to assess material properties of bone tissues. Static properties [5], dynamic properties [6] and mechanical anisotropy [7, 8] of compact bone are dependent on the age of donor [9], the osteon orientation [8, 10] and obviously bone's health [5]. Mechanical anisotropy is also shown by trabecular bone [11, 12], the loading history of the mandible during its life service strongly affects the density and the strength of trabecular bone through remodelling. Generally, the relationship between physicochemical and mechanical properties are determined by analysing the structural organization and the chemical composition [10, 13]. Explanted mandibles are also used to assess the biomechanical behaviour of the jaw during various masticatory conditions [14–16].

These experimental results are necessary to calibrate models to be used for more complex simulations of mastication [17, 18]. However, the experimental testing of natural tissues suffers problems related to the age of donors and thus bone's quality over obvious ethical reasons. In fact, the supply of cadaveric skeletal segments and the variability of the mechanical properties

*Author to whom all correspondence should be addressed.

of bone has always been a problem, both in the maxillofacial and orthopaedic fields, requiring a large number of specimens to obtain a satisfactory statistical significance in the measured data. Moreover, due to progressive degradation *in vitro*, testing over long periods of time may significantly affect the experimental measurements, leading to an improper transfer of these results to clinical trials. Therefore, attention has been paid to synthetic models as substitute for *in vitro* testing [19–22].

The finite element method (FEM) is a useful tool in analysing the stress patterns in natural and synthetic structures of complex shapes, loading and material behaviour such as the femur and the hip prostheses [23, 24] and the jaw and teeth implants [25–28]. However, the results of the FEM simulation describing the real structure behaviour have to be verified with experimental work before any conclusion can be drawn from the theoretical predictions. This approach has been adopted by a number of researchers since the numerical and experimental testing methods, used together and in parallel, are of greater value than either technique used alone [24, 29–31].

Imaging data and rapid prototyping techniques strongly simplified the manufacturing of polymeric models of temporal bone [24, 32]. Polymeric mandibular analogues (i.e. acrylic resin and polyurethanes) are used for medical teaching, training and research [19–22, 35, 36]. Although rapid prototyping through 3D printers allows an easy and fast method of reproducing solid system starting from medical images (i.e. TAC, NMR etc.), these synthetic models behave like homogeneous isotropic materials, thus the cortical and spongy bone behaviour is not distinguished. Therefore, the conclusions which may be drawn on the stability of bone and osseointegrated implants by using these experimental models of the jaw are misleading. In order to overcome this problem the outer shell of bone replicates may be reinforced with continuous fibers, reproducing the inhomogeneity and the anisotropy of structures like bone tissue [37].

In the orthopaedic field synthetic models of femur and tibia, made of glass fibre reinforced epoxy and polyurethane foam (which mimic the cortical and the spongy bone respectively), are commercially available and they are suitable for certain types of biomechanical tests [38]. However, a composite model of a human mandible, made with materials which are designed to mimic the spongy and cortical bones anisotropy, is still not available.

In a previous work glass fibre reinforced polymers have been investigated as materials for compact bone analogues. A composite mandible, consisting of a PMMA inner core and a glass fibre reinforced epoxy for the outer cortical shell, has been designed and manufactured in order to replicate the organisation of natural mandibles [37].

The composite model of the mandible can be used to replicate a variety of human mandibles since a range of mechanical properties can be easily obtained simply by changing the fibre amount of the composite shell. Moreover, a FE model of the synthetic jaw prototype is developed and calibrated on the experimental model based on local strains and arch widths measurements by loading the synthetic prototype in a suitable manner. The FE model is also used as an auxiliary tool in order to adjust the technological parameters (e.g. fibre volume ratio, orientation) to optimize the experimental model and to run simulations. In order to evaluate the accuracy in reproducing the mechanical behaviour of the natural jaw, the model is loaded according to the lateral component of the pterygoid muscles and data are compared to *in vivo* and *in vitro* literature trials.

2. Materials and methods

Composite mandibles are manufactured by using the rapid prototyping technique and the filament winding technology. The model of the jaw is derived from CT scans on a human mandible [18, 25, 34]. The inner region of the mandible (the spongy bone) is exported as an stl data file (Fig. 1(a) and (b)). The data is fed into a 3D printer (ZPRINTER® 400, CMF Marelli s.r.l., Milano, Italy) and a model of the core of the mandible is obtained through a layer by layer stratification (Fig. 1(c)). Hence, a silicone rubber mould is realised from the jaw core prototype.

In order to simulate the compact bone of the mandible arch, glass fibres epoxy prepreg (Narmco Materials, California) type 120 and type 3200 (with a laminated thickness of 127 μm and 254 μm respectively) are used in order to manufacture two types of jaw prototype with different stiffnesses.

Two plies of glass fibres epoxy prepreg are applied to the walls of the silicone rubber mould with the fibre angle of 0°, 90° with respect to the main axes of the jaw. PMMA-based self-curing bone cement (Symplex P, Howmedica®) is then injected into the mould using a syringe for hip arthoplasty. PMMA has been chosen as the core of the mandible since the Young's modulus of spongy bone in the human mandible reaches values up to 1 GPa [9, 39, 40], thus close to the one of PMMA. Moreover this polymer shows viscoelastic properties similar to those of spongy bone [37]. The synthetic jaw is then removed from the mould after 60 min and glass fibres preimpregnated with epoxy resin are helically wound around the jaw with an angle of ±45°. This composite shell simulates the compact bone layer of the mandible ramous in which osteons are mainly oriented at 45° in the ramous and at 0° in the body of the mandible [10]. The composite jaw (Fig. 1(d)) is finally coated with a poly(ethylene) shrinking film and cured at 110 °C for two h.

2.1. Mechanical testing

Ten jaw prototypes are realised and divided in two groups (A and B) according to the outer shell material type. The strain gauge technique is used to monitor local strain along the mandibular arch [22]. Strain gauge rosettes (CEA-13-062-UR-120 measurements group, Inc. Raleigh, North Carolina, USA) are applied on the inner and outer surface of the jaw as showed in Fig. 2. The scanner system 5100B® (Vishay

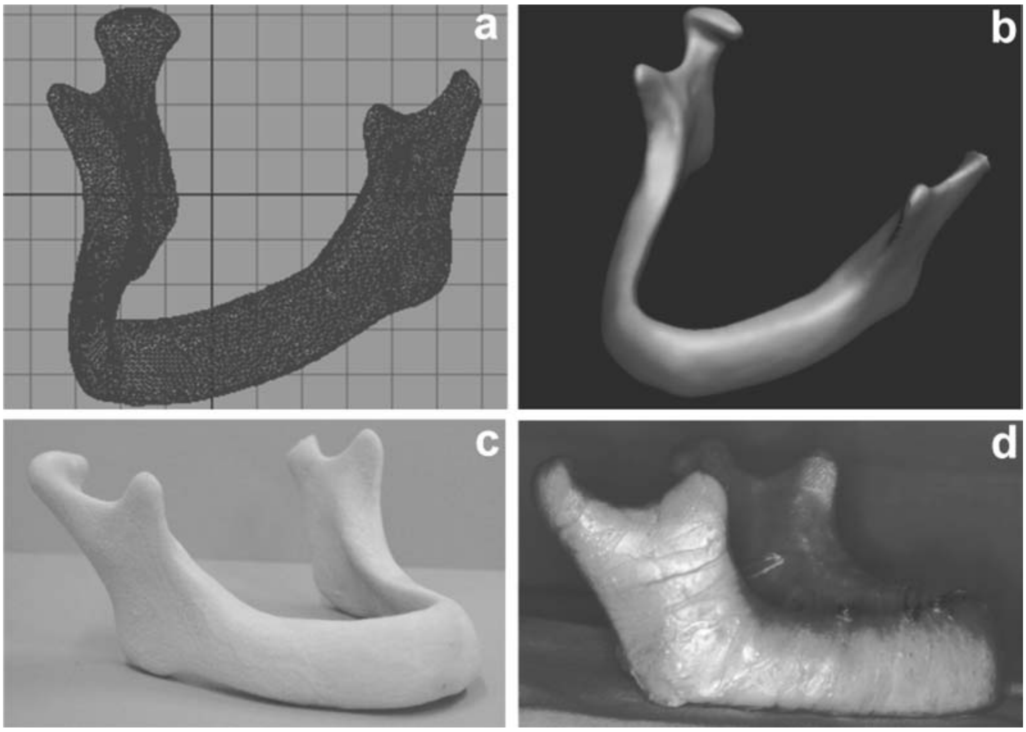

Figure 1 The data file of the inner core of the mandible (a) and its 3D rendering (b). The model of the core of the mandible obtained through a layer by layer stratification (c). The mandible model made of a PMMA inner core and a composite outer shell (d), the poly(ethylene) shrinking film is also shown.

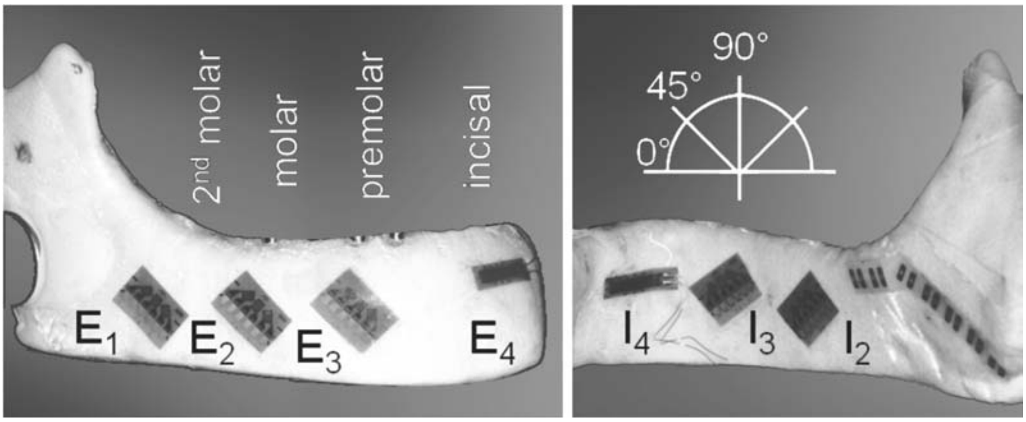

Figure 2 Instrumented composite mandible. Strain gauges are applied on the outer (E1–E4 gauges) and inner surfaces (I2–I4 gauges) of the experimental model at different sites. Rosettes are oriented in order to directly measure the strain along the mandible arch axis (0° direction).

Micro-measurements Raleigh, North Carolina, USA) is used to acquire the load displacement data and local strain gauge signals at a rate of 10 pt/s. In order to evaluate the arch width change as the load is increased, the optical displacement measurement VE5000 (Trio Sistemi e Misure, Brescia, Italy) controlled by the LabView 7.0 software package (National Instruments Corp., Austin, USA) is employed.

The static behaviour of the composite jaw is analysed by laterally loading the device in the condyle (case A) or in the gonion (case B) regions of the sagittal mandible ramous (Fig. 3(a)), this loading condition being the lateral component of the pterygoid muscles [41, 42].

The Instron 4204 dynamometer with a load cell of 100 N is used to perform the mechanical testing on systems represented by a distally supported or cantilevered bridges at a rate of 1 mm/min.

2.2. FEM analysis

The inner region of the mandible (the spongy bone) obtained with the CT scan is exported as an Iges file to the commercial FE software, ANSYS 6.0 (ANSYS Inc. Canonsburg, PA, USA). The profile of each section resulting by the scanning has been approximated with cubic splines. This method is very effective to obtain a smooth the external surface. The corresponding sections were then joined together (Fig. 4(a)) to create continuous inner volumes. All the volumes could then finally be meshed with linear brick unlayered elements as illustrated in Fig. 4(b).

In order to simulate the outer windings of the composite mandible, a layered brick element with 8 nodes is used. A coordinate system for all of the layers is created: the X axis is directed along the axial direction, while the Z axis is normal to each layer. In this way it is possible to set the number of layers and to specify the

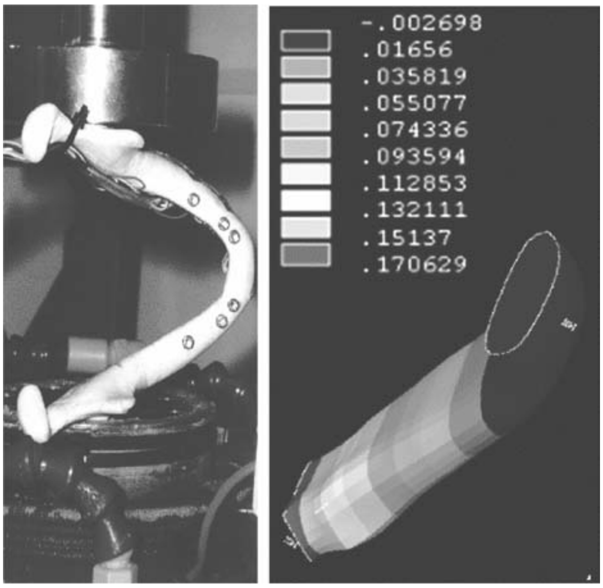

Figure 3 Mechanical set up of the composite mandible loaded in the gonion regions (a) and the simulation results representing the displacement field related to one half of the mandible (b).

orientation, thickness, and the constituent material for each layer. The guided finite element mesh generation approach (Fig. 4) is preferred to meshing with tetrahedral elements to gain control over the orientation of the material properties more easily. Moreover, meshing with tetrahedral elements may produce brick element degeneration, which can lead to fictitious stress concentrations. The meshed volumes resulting from this study did not show degenerating elements.

Two layers are simulated, the first one with the fibres oriented at 0° and 90°, like the Narmco prepreg, and the second one with the fibres oriented at ±45° as in the fibres wound following the release from the mould (Fig. 4(c)).

The PMMA inner core has been assumed to be linear elastic and isotropic, while the glass reinforced epoxy resin is assumed to be an orthotropic material. The material properties used for the simulation are listed in Table I.

TABLE I Material properties used for the simulation

Cortical bone—reinforced fibers Composite	Ex = Ey	21000 MPa
	Ez	1000 MPa
	Gxz = Gyz	7000 MPa
	Gxy	333 MPa
	ν_{xy}	0.05
	$\nu_{xz} = \nu_{yz}$	0.3
Cancellous bone—PMMA	E	2500 MPa
	ν	0.3

In order to reproduce the experiments performed on the synthetic jaw, a 10 N load is applied in the gonion region as described in Fig. 3(a). Given the symmetry of the mandible and of the loading about the sagittal plane, the study is performed only on a half mandible.

3. Results

The geometrical properties of the mandible models are depicted and listed in Fig. 1 and Table II.

The stiffness of group A mandibles, loaded in the gonion region through the sagittal mandible ramous (Fig. 3(a)), is found to be 35 N/mm (±1.5 N/m). This value is in close agreement with the stiffness measured through the FEM analysis (33 N/mm). Particularly the convergence of the gonion region at 10 N is 0.28 mm (±0.01 mm)) and this value is consistent with the displacement simulation results of the gonion region in Fig. 3(b) (FEM displacement results being half of the width change) related to one half of the mandible. Group B mandibles have reported a stiffness of 67 N/mm (±2 N/m). The stress-strain behaviour of the synthetic jaws are elastic and linear for both the external and the inner strain gauges at least up to 40 N.

A close agreement is found between the mean values of the experimental and theoretical strains due to a 10 N load.

Fig. 5 compares the experimental and theoretical results for a 40 N loading along the 0° direction (that is the mandible's axis) in both the labial and lingual regions

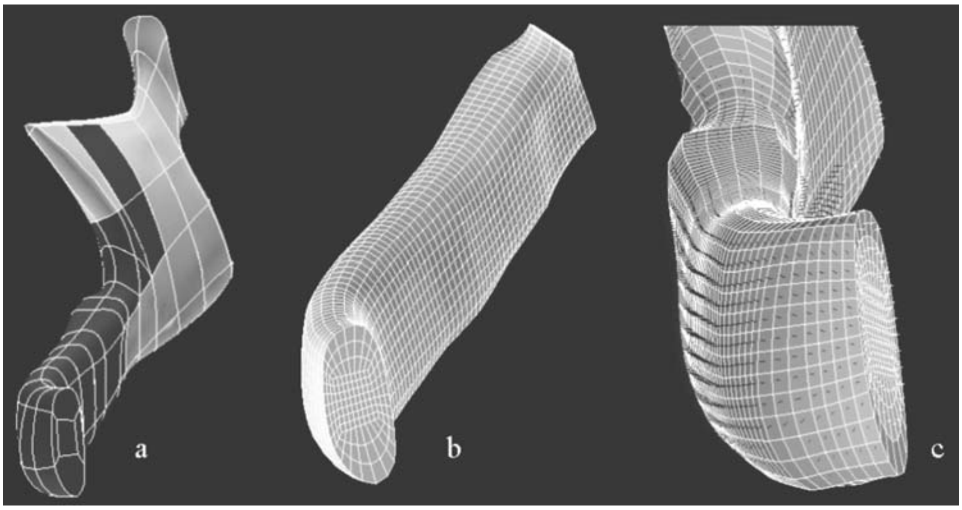

Figure 4 FEM model of the composite mandible: continuous inner volumes (a) joining the corresponding vertexes of rectangular perimeters inside each section of the mandibular arch; (b) the mapped meshing of the inner homogenous part of the mandible through isotropic unlayered solid elements; (c) layers of the outer shell oriented along the axial direction.

TABLE II Dimensions and properties of the mandible prototypes

Length (mm)			Width (mm)		Thickness (mm)		
Condyle symphisy	Condyle gonion	lower 1st molars	Sagittal ramous	Mandible arch	Sagittal ramus	Mandible arch	Weight (g)
120	63	54	32	24	16	15	95 (±3)

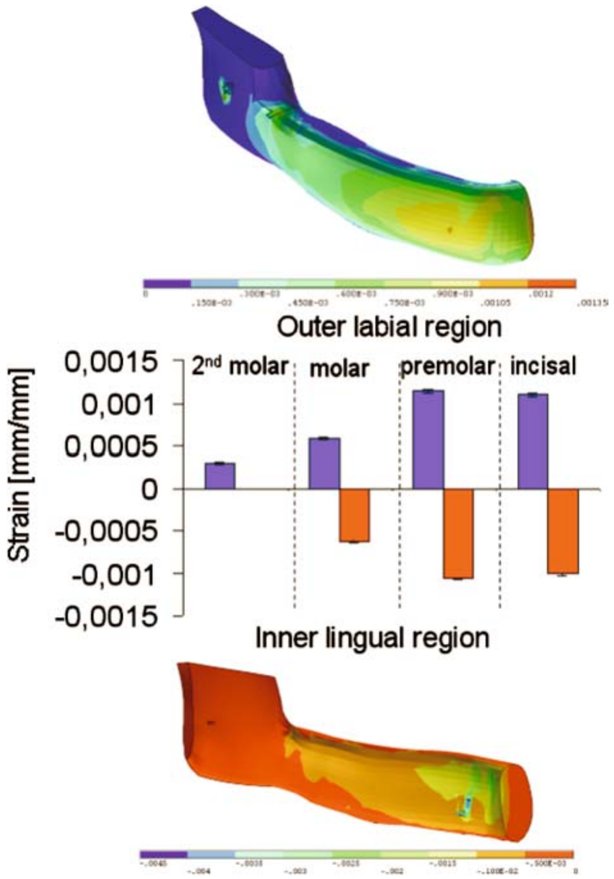

Figure 5 Experimental and theoretical results for a 40 N loading along the 0° direction in both the labial and lingual regions as a function of the mandible site. On the labial surface of the mandible (outer region) the strain is positive and reaches a maximum in the premolar region. In the lingual region (inner surface of the mandible) the strain is negative and increases in the molar incisal direction reaching a maximum in the incisal region. The FEM results also show that higher values of strain are found in the premolar region of the labial surface and close to the symphysis region of the lingual surface.

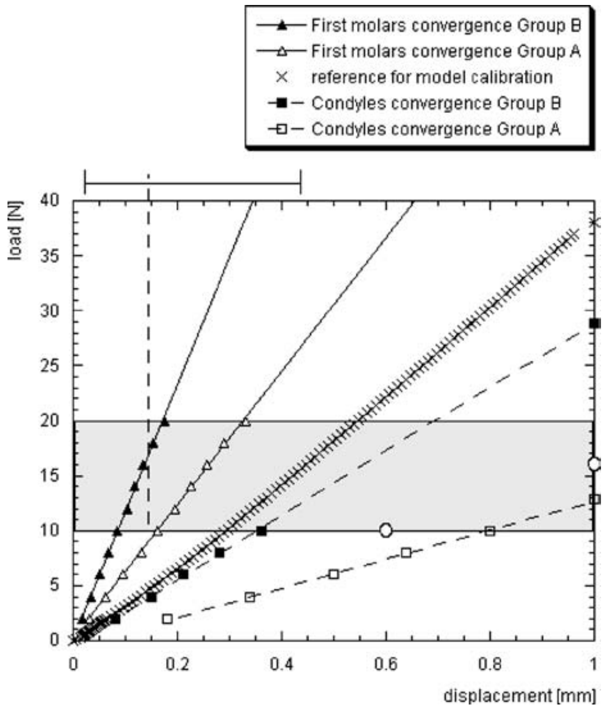

Figure 6 Condyles (square symbols) and first molars (triangle symbols) convergence according to case A loading condition, empty and full symbols being the results on group A and group B samples respectively. The reference curve refers to the gonion convergence of group A mandibles loaded as case B and the resulting model calibration. The circles represent condyles convergence of a human mandible [19, 24] loaded in the same manner of our experiment. On the top of the *X* axis the range of first molar convergence of young human mandibles during mouth opening and closing movements [30] and the mean value (dashed vertical line) are also reported.

as a function of the mandible site. On the labial surface of the mandible (outer region) the strain is positive and reaches a maximum in the premolar region. In the lingual region (inner surface of the mandible) the strain is negative and increases in the molar incisal direction reaching a maximum in the incisal region. The FEM results also show that higher values of strain are found in the premolar region of the labial surface and close to the symphysis region of the lingual surface, as depicted in Fig. 5.

Fig. 6 compares the composite model results to literature data. The reference curve refers to case B loading condition and the theoretical model calibration. As the mandible is loaded according to case A, condyles convergence and first molars convergence can be directly derived. The circles represent the results of condyles convergence of a human mandible loaded in the same manner of our experiment [25, 35]. On the top of the *X* axis the range of first molar convergence and the mean value (dashed vertical line), measured *in vivo* on young human mandibles during mouth opening and closing movements is also reported [43].

4. Discussion

Polymeric composites are among the most suitable materials for manufacturing tissue replicates. The architecture of almost all natural tissues suggests that a continuous fibre reinforcement design is a powerful tool in order to optimise the mechanical function. Based on the same material design approach it has been possible to engineer composites to reflect the mechanical anisotropy of natural connective tissues more accurately [37]. The carbon and glass fibre reinforcement is a suitable option to design composites for prostheses with a stem fitting inside a canal (i.e. hip prostheses and dental posts) in order to match stiffness properties similar to the surrounding hard tissue [37, 44]. On the other hand, in the case of bone replicates several types of fibre reinforcement are available in order to achieve the stiffness of the cortical bone. A glass reinforcement is particularly attractive since it offers the possibility to manufacture relatively low cost synthetic bone substitutes for designing and testing purposes [38, 45]. Moreover, the continuous winding approach seems to be the most appropriate technology in order to keep a fine control of the material anisotropy [37, 44]. The experimental jaw model presented in this report shows that a wide range of properties may be covered by simply changing the outer shell material density, thus keeping fibre angle deposition close to the average osteon orientation in the human mandible [9, 10]. Furthermore, the weight of this synthetic mandible (95 g) is close to the natural one [46].

As a synthetic bone model is developed, an experimental protocol needs to be defined in order to test its mechanical performance [38]. Case A and case B loading conditions represent a simple testing method to assess the composite mandible mechanical behaviour. The lateral loading of the device through the condyles or through the gonion regions (Fig. 3(a)) are implemented in order to characterise the whole mandible or the coronal mandible arch respectively. Case A loading condition reproduces the controlateral component of the pterygoid muscles which act in the lateral medial direction of the frontal mandible view [27, 47–49]. In Case B the load is applied in the gonion region of the sagittal ramous of the mandible, almost close to the centroid of the lower jaw around which relevant forces and torques due to muscles and joint generate the mandible movement [50]. Case B loading condition is a quick attractive protocol not only to assess the mandible arch performance using a single axis dynamometer, but also to check and optimise implant-arch bridge and orthodontic designs which will be the topic of future researches.

Strain gauges are used to locally measure the strain at selected locations. Even if strain gauges and optical methods are able to accurately measure the strains and the displacements at the surface [20–22], it is impossible to obtain information on the internal strain and stress distribution. Thus, a simulation using FEM based on the experimental model can be very convenient in order to complete the stress analysis. FEM simulation, though, is unable to study the phenomenon by itself, since it has to be checked with experimental work. For this reason, the development of the theoretical model is validated through comparison between the theoretical predictions and the experimental findings (Figs. 3–5).

The effects of the lateral component of pterygoid muscles on the mandible deformations during the mouth open close movements are well recognised [35–48]. In fact, the decrease of the arch width at the first molar level has been detected and precisely measured *in vivo* during the mouth opening and closing [43]. Moreover, pterygoid muscles are crucial also during clenching activity since they increase the jaw effort through a fine control of mandible movements thus allowing the development of higher forces by the other masticatory muscles [51].

While displacements of skeletal elements can be easily determined through imaging and displacement sensors techniques [43, 48], the forces developed by the muscles of the masticatory apparatus and temporomandibular joints cannot be directly measured *in vivo*. Nevertheless, the assessment of both the force and displacement are essential for a proper knowledge of the mandible biomechanics [52]. Thus, data on the forces generated by muscles are mainly based on theoretical models which estimate the force depending on the muscle architecture, the sarcomers length [47, 51–53] and electromyographic activity [51, 54]. The maximum tension which can be developed by the pterygoid muscles has been reported to be 170 N and 67 N for the medial and lateral pterygoid muscle respectively [54, 55], while maximum values of 50 N and 100 N have been distinguished for the superior head and the inferior head of the lateral pterygoid muscle [41]. Like all the other muscle-tendons units, the output force can be described as a parabolic function of the muscle length [47]. However, only the horizontal lateral component, in the frontal plane of the mandible, is responsible for the arch width decrease and it can be derived according to the average direction of the muscle lines of action [27, 47, 53]. A lateral load between 10 N and 20 N is thus obtained and used to estimate the mandible prototypes deformations (Fig. 6).

The mandible loaded as in case A shows that condyles displacement ranges between 0.36 mm and 1.53 mm. Particularly, the Group A mandibles show a linear elastic behaviour which is very close to the results obtained on a human mandible loaded in the same manner [25] with 10 N and *in vivo* findings on human edentulous mandible [35] loaded with a lateral component of 16 N. Thus, our model also links clinical trials [43] to another theoretical model of the human mandible [25]. The range of deformations predicted and measured at the first molars level of our mandible model (0.083–0.327 mm) due to a lateral load of 10–20 N is consistent with the displacement results obtained *in vivo* during mouth opening of subjects not affected by mandibular disorder with an average age of 23 years [43].

5. Conclusion

A composite mandible made of a PMMA core and a glass reinforced outer shell is manufactured and characterised through mechanical tests assisted by the FE analysis. The symmetry of the mandible suggests that a suitable testing protocol to assess the biomechanical properties of the mandible replicate can be obtained by loading the mandible laterally in the gonion region. A close agreement is found between experimental data and theoretical predictions.

These results are consistent with literature data: in this loading condition maximum value of stress and strain take place in the premolar-incisal region, which is the region of maximum curvature. Thus, in clinical trials, these are the regions where high stress concentrations occur as a result of the lateral component of pterygoid muscles activity.

This novel synthetic system coupled with the FE model constitutes an experimental-theoretical model suitable to investigate the biomechanical effects on mandibular bone of oral rehabilitations.

Acknowledgement

The financial support of CRdC of regione Campania WP1C and the Cluster C/26 project "Biomateriali in funzione applicativa" are greatfully acknowledged.

The authors also wish to thank Mr. Rodolfo Morra, Dr. Antonio Gloria, Dr. Saverio Maietta of the Department of Material and Production Engineering, University of Naples Federico II for the mechanical measurements and Francesco Amoroso of Sistemi Compositi SpA for the material supply.

References

1. H. PLENK, *J. Biomed. Mater. Res. Part B: Appl. Biomat.* **43** (1998) 350.
2. I. BALTAG, K. WATANABE, H. KUSAKARI, N. TAGUCHI, O. MIYAKAWA, M. KOBAYASHI and N. ITO, *ibid.* **53** (2000) 76.
3. S. LEWIS, S. PAREL and R. FAULKER, *Int. J. Oral. Maxillofac. Implants* **10** (1995) 319.
4. T. M. WANG, L. J. LEU, J. WANG and L. D. LIN, *ibid.* **17** (2002) 231.
5. C. L. SCHWARTZ-DABNEY and P. C. DECHOW, *J. Dent. Res.* **81** (2002) 613.
6. V. VITINS, M. DOBELIS, J. MIDDLETON, G. LIMBERT and I. KNETS, *Comput. Methods Biomech. Biomed. Engng.* **6** (2003) 299.
7. T. NOMURA, J. L. KATZ, M. P. POWERS and C. SAITO, *J. Biomed. Mater. Res. Part B: Appl. Biomat.* **73** (2005) 29.
8. S. LETTRY, B. B. SEEDHOM, E. BERRY and M. CUPPONE, *Bone* **32** (2003) 35.
9. T. HARA, M. TAKIZAWA, T. SATO and Y. IDE, *Bull Tokyo Dent. Coll.* **39** (1998) 47.
10. Y. TAMATSU, K. KAIMOTO, M. ARAI and Y. IDE, *ibid.* **37** (1996) 93.
11. A. M. O'MAHONY, J. L. WILLIAMS, J. O. KATZ and P. SPENCER, *Clin. Oral. Impl. Res.* **11** (2000) 415.
12. E. B. GIESEN, M. DING, M. DALSTRA and T. M. VAN EIJDEN, *Clinical Biomechanics* **18** (2003) 358.
13. S. S. KOHLES and D. A. MARTINEZ. *J. Biomed. Mater. Res.* **49** (2000) 479.
14. D. VOLLMER, U. MEYER, U. JOOS, A. VEGH and J. PIFFKO, *Journal of Cranio-Maxillofacial Surgery* **28** (2000) 91.
15. C. MEYER, J. L. KAHN, P. BOUTEMI and A. WILK, *ibid.* **30** (2002) 160.
16. D. RODRIGUEZ, V. MORENO, M. GALLAS, M. T. ABLEIRA and D. SUAREZ, *Med. Eng. Phys.* **26** (2004) 371.
17. R. T. HART, V. V. HENNEBEL, N. THONGPREDA, W. C. VANBUSKIRK and R. C. ANDERSON, *J. Biomechanics* **25** (1992) 261.
18. A. APICELLA, E. MASI, L. NICOLAIS, F. ZARONE, N. DE ROSA and G. VALLETTA, *J. Mat. Sci.:Materials in Medicine* **9** (1998) 191.
19. K. B. TAN and J. I. NICHOLLS, *Int. J. Oral Maxillofac. Implants* **17** (2002) 175.
20. J. A. PORTER JR, V. C. PETROPOULOS and J. B. BRUNSKI, *ibid.* **17** (2002) 651.
21. M. KARL, W. WINTER, T. D. TAYLOR and S. M. HECKMANN, *ibid.* **19** (2004) 30.
22. M. M. NACONECY, E. R. TEIXEIRA, R. S. SHINKAI, L. C. FRASCA and A. CERVIERI, *ibid.* **19** (2004) 192.
23. A. APICELLA, A. LIGUORI, E. MASI and L. NICOLAIS, in "Total hip replacement". (Sheffield Academis Press, Sheffield, 1994) 323.
24. M. AKAY and N. ASLAN, *J. Biomed. Mater. Res.* **31** (1996) 167.
25. F. ZARONE, A. APICELLA, L. NICOLAIS, R. AVERSA and R. SORRENTINO, *Clin. Or. Implants Res.* **14** (2003) 103.
26. K. AKCA and H. IPLIKCIOGLU, *Int. J. Oral Maxillofac. Implants* **16** (2001) 722.
27. M. CRUZ, T. WASSALL, E. M. TOLEDO, L. P. BARRA and A. C. LEMONGE, *Int. J. Oral Maxillofac. Implants* **18** (2003) 675.
28. M. SUTPIDELER, S. E. ECKERT, M. ZOBITZ and K. N. AN, *Int. J. Oral Maxillofac. Implants* **19** (2004) 819.
29. R. HUISKES, J. D. JANSSEN and T. J. SLOFF, in "Mechanical properties of bone" (Cowin SC, ASME Editor, New York, 1981) p. 221.
30. R. DE SANTIS, D. PRISCO, A. APICELLA, L. AMBROSIO, S. RENGO and L. NICOLAIS, *J. Mat. Sci.: Materials in Medicine* **4** (2000) 201.
31. H. LINDSTROM and H. PREISKEL, *Int. J. Oral. Maxillofac. Implants* **16** (2001) 34.
32. E. BERRY, J. M. BROWN, M. CONNELL, C. M. CRAVEN, N. D. EFFORD, A. RADDJENOVIC and M. A. SMITH, *Med. Eng. Phys.* **19** (1997) 90.
33. P. S. D'URSO, T. M. BARKER, W. J. EARWAKER, L. J. BRUCE, R. L. ATKINSON, M. W. LANIGAN, J. F. ARVIER and D. J. EFFENEY, *J. Craniomaxillofacial. Surg.* **27** (1999) 30.
34. M. NAITOH, A. KATSUMATA, S. MITSUYA, H. KAMEMOTO and E. ARIJI, *Int. J. Oral Maxillofac. Implants* **19** (2004) 239.
35. J. A. HOBKIRK, *ibid.* **6** (1991) 319.
36. P. POTAMIANOS, A. A. AMIS, A. J. FORESTER, M. MCGURK and M. BIRCHER, *Proc. Inst. Mech. Eng.* **H212** (1998) 383.
37. R. DE SANTIS, F. SARRACINO, F. MOLLICA, P. A. NETTI, L. AMBROSIO and L. NICOLAIS, *Comp. Sci. Tech.* **64** (2004) 861.
38. L. CRISTOFOLINI, M. VICECONTI, A. CAPPELLO and A. TONI, *J. Biomechanics* **29** (1996) 525.
39. G. LEWIS, *J. Biomed. Mater. Res. Part B: Appl. Biomat.* **38** (1997) 155.
40. C. E. MISCH, Z. QU and M. W. BIDEZ, *J. Oral Maxillofacial Surg.* **57** (1999) 700.
41. J. W. OSBORNE, *Arch. Or. Biol.* **40** (1995) 1099.
42. J. H. KOOLSTRA and T. M. G. J. VAN EIJDEN, *J. Biomechanics* **30** (1997) 883.
43. D. C. CHEN, Y. L. LAI, L. Y. CHI and S. Y. LEE, *J. Dent.* **28** (2000) 583.
44. S. RAMAKRISHNA, J. MAYER, E. WINTERMANTEL and K. W. LEONG, *Comp. Sci. Tech.* **1** (2001) 1189.
45. C. I. VALLO, *J. Biomed. Mater. Res. Part B: Appl. Biomat.* **53** (2000) 717.
46. F. ZHANG, C. C. PECK and A. G. HANNAM, *J. Biomechanics* **35** (2002) 975.
47. T. M. G. J. VAN EIJDEN, J. H. KOOLSTRA and P. BRUGMAN, *J. Dent. Res.* **74** (1995) 1489.
48. G. M. MURRAY, I. PHANACHET, M. HUPALO and K. WANIGARATNE, *Arch. Or. Biol.* **44** (1999) 671.
49. I. PHANACHET and G. M. MURRAY, *ibid.* **45** (2000) 517.
50. J. H. KOOLSTRA and T. M. G. J. VAN EIJDEN, *J. Dent. Res.* **74** (1995) 1564.
51. I. PHANACHET, K. WANIGARATNE, T. WHITTLE, S. UCHIDA, S. PEECEEYEN and G. M. A. MURRAY, *J. Neur. Meth.* **105** (2001) 201.
52. J. H. KOOLSTRA, *J. Dent. Res.* **82** (2003) 672.
53. B. MAY, S. SAHA and M. SALTZMAN, *Clin. Biomech.* **16** (2001) 489.
54. G. E. J. LANGENBACH and A. G. HANNAM, *Arch. Or. Biol.* **44** (1999) 557.
55. C. C. PECK, G. E. J. LANGENBACH and A. G. HANNAM, *ibid.* **45** (2000) 963.

Received 20 July
and accepted 19 August 2005