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Reverse engineering of mandible and prosthetic framework: Effect of titanium implants in conjunction with titanium milled full arch bridge prostheses on the biomechanics of the mandible

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ABSTRACT

This study aimed at investigating the effects of titanium implants and different configurations of full-arch prostheses on the biomechanics of edentulous mandibles. Reverse engineered, composite, anisotropic, edentulous mandibles made of a poly(methylmethacrylate) core and a glass fibre reinforced outer shell were rapid prototyped and instrumented with strain gauges. Brånemark implants RP platforms in conjunction with titanium Procera one-piece or two-piece bridges were used to simulate oral rehabilitations. A lateral load through the gonion regions was used to test the biomechanical effects of the rehabilitations. In addition, strains due to misfit of the one-piece titanium bridge were compared to those produced by one-piece cast gold bridges. Milled titanium bridges had a better fit than cast gold bridges. The stress distribution in mandibular bone rehabilitated with a one-piece bridge was more perturbed than that observed with a two-piece bridge. In particular the former induced a stress concentration and stress shielding in the molar and symphysis regions, while for the latter design these stresses were strongly reduced. In conclusion, prosthetic frameworks changed the biomechanics of the mandible as a result of both their design and manufacturing technology.

1. Introduction

Osseointegrated implants in conjunction with full-arch prostheses are being used increasingly in oral rehabilitation to restore the physiological functions of edentulous patients. The biomechanics of a mandible rehabilitated with implant-supported full-arch bridges is different from that of a healthy mandible: implants are rigidly connected together by the prosthesis, lacking any shock absorbing capacity at the bone interface (Ishigaki et al., 2003; Natali and Pavan, 2003). When a one-piece full-arch prosthesis is used to rehabilitate edentulous mandibles, additional implants placed posterior to the mental foramen are at a higher risk of failure compared to their anterior counterparts (Miyamoto et al., 2003) probably due to mandible deformation. Previous biomechanical studies reported that an implant supported full-arch rehabilitation is affected by the deformation of the mandible already in the simple case of mouth opening and closing (Apicella et al., 1998; Koolstra and van Eijden, 1995; Zarone et al., 2003). During this activity, a lateral component of the pterygoid muscle determines an arch width decrease by exercising an estimated load between 10 N and 20 N (Chen et al., 2000; Koolstra, 2003; Langenbach and Hannam, 1999; Murray et al., 1999; Phanachet et al., 2001). As small as these loads might seem the resulting mandible deformations are entirely transferred to the peri-implant bone where, due to the splinting effect of the prosthesis and the lack of any damping ability, they turn out in high stress concentration. Therefore, mandible deformation is of concern in implant dentistry since it is very frequent (Peck et al., 2000) and its effect sums up with that of the misfit that is systematically observed at one-piece long-span prosthesis (Torsello et al., 2008). Any prosthetic misfit induces potentially detrimental stress states in the peri-implant bone although the noxious effect of such misfit has not been clinically quantified yet (Natali et al., 2006).

The realisation through a reverse engineering approach of solid mandible models, recently introduced by De Santis et al. (2004), is promising to improve the knowledge in implants biomechanics as,

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contrary to other theoretical models reported in the literature (Porter et al., 2002; Sutpideler et al., 2004; Tan and Nicholls, 2002; Zarone et al., 2003), it allows reproduction of human jaw anisotropy (De Santis et al., 2007; Schwartz-Dabney and Dechow, 2003). Briefly, a customised 3D solid model based on radiographic imaging of a patient mandible is reproduced through rapid prototyping of an inner poly(methylmethacrylate) (PMMA) core completed with a layer of suitably oriented synthetic fibres (De Santis et al., 2004). Here these mandibular models will be used to investigate the effects of different configurations of implant supported full-arch prostheses on mandible biomechanics, the aim being to compare the bone strain induced when fitting either computer-aided design/computer-assisted manufacturing (CAD/CAM) milled titanium or cast gold alloy frameworks on mandibular implants and to analyse the stiffness of mandibles rehabilitated with one-piece or two-piece implant-supported CAD/CAM milled titanium frameworks during simulated activity of the pterygoid muscles in the phases of mouth opening and closing.

2. Materials and methods

15 composite edentulous mandibles were rapid-prototyped by using a 3D printing technique in conjunction with the composite materials technology, as described in a previous work (De Santis et al., 2004). The inner core of the composite mandible consisted of a PMMA based self-curing bone cement (Symplex P, Howmedica® Stryker, Kalamazoo, Michigan, USA), with mechanical properties similar to spongy bone (De Santis et al., 2007). Hence, trabecular bone was considered as an isotropic material and it was replicated with PMMA based bone cement. Young's modulus of this bone cement is 2.6 GPa (De Santis et al., 2003) and this value is very close to the Young's modulus of 2.2 GPa measured for trabecular bone in the mandible symphysis and along the bucco-lingual direction (O'Mahony et al., 2000).

The outer shell of the mandible model consisted of glass fibre reinforced epoxy with a laminated thickness of 127 μm (Prepreg type 120, BASF Structurals Materials Inc, Narmco Division, Anaheim, California, USA). In order to simulate the compact bone anisotropy of the mandible arch, fibres were oriented at angles of 0°, 90° with respect to the axis of the mandible corpus while in the ramus they were oriented at angles of $\pm 45°$ (Schwartz-Dabney and Dechow, 2003).

In order to validate the composite mandible model, experimental testing was carried out by loading composite mandibles through the condyles. This loading condition reflects the loading configuration adopted by Hobkirk and Schwab (1991) and Zarone et al. (2003).

Mandibles were then divided into three groups, namely control group, group A and group B, each composed of 5 specimens. Mandibles in the control group were not modified further. Conversely, in each mandible of groups A and B, six parallel implant sites were drilled in canine, first premolar and first molar areas with the aid of a parallelometer (Cendres+Metaux, Biel, Switzerland). In such sites, dental implants (8.5 mm ⌀3.75 mm Brånemark System® RP, Nobel Biocare, Goteborg, Sweden) were cemented using the same PMMA bone cement as above (Fig. 1a and b). A regular viscosity polyether (Permadyne, 3M ESPE, St Paul, Minnesota, USA), mixed through an appropriate dispenser (Pentamix 2, 3M ESPE), was used for implant level impressions of all the mandibles. Once a model was obtained from each impression, an acrylic resin replica of the final framework was fabricated. The replica was then laser scanned according to the "All in one" Procera workflow (Nobel Biocare) to finally obtain 10 identical titanium frameworks. 5 frameworks were left unmodified as one-piece appliances and were assigned to group A while the remaining 5 frameworks were cut into two halves between the central incisors and assigned to group B. Furthermore, 5 additional cast gold frameworks, matching the outline of the resin replica used for Procera bridges were manufactured using conventional techniques. These prostheses were connected to group A mandibles, alternately to Procera titanium bridges, to compare bone strains eventually due to the misfit of the two frameworks. All the prostheses were tightened with a wrench according to manufacturer's indications. To monitor local strain along the mandibular arch, strain gauges (CEA-13-062-UR-120, Vishay Micro-Measurements, Raleigh, North Carolina, USA) (Fig. 1) were bonded to the vestibular and lingual surfaces of each mandible in incisor, premolar and molar areas. A data acquisition system (5100B® Vishay Micro-measurements, Raleigh, North Carolina, USA) was used to record the load–displacement data and local strain gauge signals at a rate of 10 pt/s (Fig. 2a).

The first experiment was run by recording the bone strain occurring to group A mandibles after alternately screw-tightening Procera titanium or cast gold frameworks to the implants.

The second experiment was run to record the stiffness of control group, group A and group B mandibles when they were symmetrically loaded along the occlusal plane as a cantilevered bridge system as depicted in Fig. 2. This loading condition approximated the lateral component of the action of the pterygoid muscles. A dynamometer (Instron 5566, Instron. Bucks, UK) was used to perform mechanical testing at a crosshead speed of 1 mm/min up to a maximum loading of 40 N. ANOVA at a significance level of 0.01, followed by the Tukey post-hoc test, was used to compare measurements among groups.

3. Results

The stiffness of the experimental mandible model loaded through the condyles was 14.2 N/mm ($\pm$1.3 N/mm). The distance between the condyles reduced by 1 mm at a pterygoid muscles

Fig. 1. Anisotropic mandible model instrumented with strain gauges and rosettes: a) vestibular prospective showing implants positioning into the mandible, b) lingual prospective, c) vestibular prospective showing the mandible rehabilitated with titanium full arch bridge, and d) lingual prospective showing inner strain gauge and rosettes.

Fig. 2. Experimental set-up for the in vitro testing of mandible.

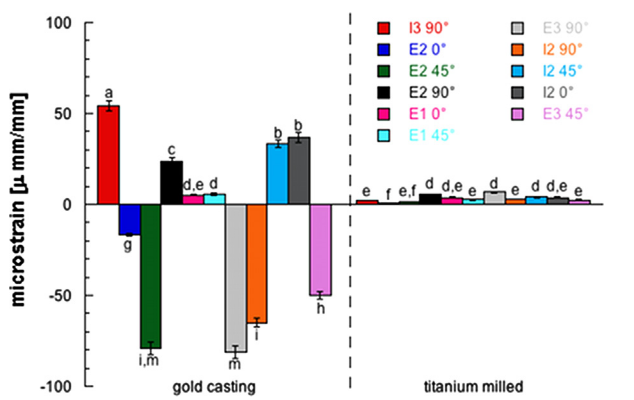

Fig. 3. Strains recorded on the labial and vestibular surfaces of the mandible in the molar region (rosettes I2 and E2, respectively), on the labial and vestibular surface of the premolar region (rosettes I3 and E3, respectively) and on the vestibular surface of the gonion region (rosette E1) when alternatively fitting Procera CAD/CAM titanium and cast gold frameworks on mandibles of the group A. Negligible strains were recorded for group A; hence an almost passive fit was recognised. Statistical differences among measurements are indicated by different letter codes. Data are graphically reported as mean value, and bars represent the standard deviation.

load of 16 N (Hobkirk and Schwab, 1991). Consequently, the *in vivo* stiffness of the mandible was 16 N/mm. On the other hand, a condyle convergence of 0.6 mm for a mandible model loaded with 10 N through the condyles has been measured (Zarone et al., 2003). Consequently, the *in vitro* stiffness of the mandible was 16.7 N/mm. Therefore the stiffness that we measured for the composite mandible model is very close to the literature data, thus validating the experimental mandible model.

Strains recorded in different directions with respect to the mandible axis on the labial and vestibular surfaces of the mandible in the molar region (rosettes I2 and E2, respectively), on the labial and vestibular surface of the premolar region (rosettes I3 and E3, respectively) and on the vestibular surface of the gonion region (rosette E1) when alternatively fitting CAD/CAM titanium and cast gold frameworks on mandibles of the group A are shown in Fig. 3. Negligible strains were recorded for group A; hence an almost passive fit was recognised.

With regard to the stiffness of mandibles in the control group, in group A and B a linear loading/deformation trend was found when analysing the mandibles up to the maximum tested load of 40 N (Fig. 4a). Stiffness, as calculated from the loading/deformation slope, was 35 N/mm ($\pm$1.5 N/mm), 43 N/mm ($\pm$1.9 N/mm) and 39 N/mm ($\pm$1.4 N/mm). A significant difference ($p<0.01$) was found between the stiffness of control group and group A mandibles.

Along the mandibular axis, in the molar, premolar and incisor regions, positive and negative strains were measured on the vestibular and lingual side, respectively (Fig. 4b). Hence, all mandibles underwent a tension and a compression stress state along the vestibular and lingual side, respectively. For group A, significantly lower strains resulted in the incisor and premolar regions compared to the control group ($p<0.01$). Conversely, in the molar region, strains were higher for group A compared to the control group ($p<0.01$).

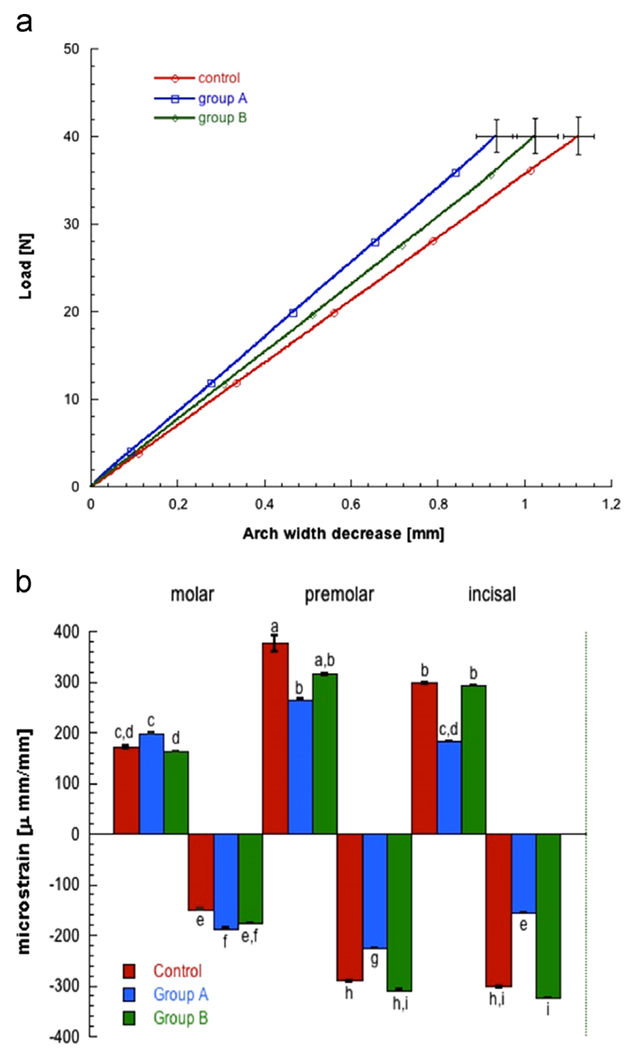

Fig. 4. (a) Stiffness of mandibles in the control group (edentulous), group A (one-piece Procera bridge) and group B (two-piece Procera bridge). Group A stiffness was significantly higher than that of the control group. (b) Strains along the mandibular axis for control group (edentulous), group A (one-piece Procera bridge) and group B (two-piece Procera bridge) measured in the molar, premolar and incisal regions. All mandibles undergo a tension and a compression state of stress along the mandible arch on the vestibular and lingual sides, respectively. Stress shielding in the premolar–incisal region due to the full arch bridge prosthesis (group A) can be recognised. Statistical differences among measurements are indicated by different letter codes. Data are graphically reported as mean values, and bars represent the standard deviation.

No significant differences were measured between the control group and group B.

4. Discussion

Engineering of dental implants could enhance the understanding of the biomechanical aspects involved in the design of an implant-supported restoration. A reverse engineering approach has been used to create customised theoretical and experimental simulations of the rehabilitated mandible (De Santis et al., 2004, 2005; Koolstra and van Eijden, 1995; Zarone et al., 2003). While the geometry of these models is easily derived from x-ray, tomography or magnetic resonance imaging, their mechanical properties still need validation through experimental testing. Unfortunately, mechanical testing of human tissues is affected by large variability in bone quality of human jaw samples (Ulm et al., 1997). Large sample size would be necessary to statistically overcome variability but, due to ethical reason, rarely the gathering of such a sample size is possible and progressive degradation may also significantly affect the experimental measurements during *ex-vivo* testing, thus causing the transfer of improper results to clinical trials.

Although mechanically different from bone, polymers and dental stone have been widely used as an alternative to the *ex-vivo* approach to replicate human mandibles for *in-vitro* validation studies (Karl et al., 2004; Naconecy et al., 2004; Porter et al., 2002; Tan and Nicholls, 2002). Similarly polyurethane resin mandibles, markedly isotropic and not customisable by reverse engineering of digitally acquired anatomical data sets, have also been used for biomechanical testing (Madsen and Haug, 2006).

In contrast, composite materials featuring continuous fibre reinforcement design might be valid candidates to replicate the mechanical properties of natural hard tissues. In fact, by controlling the fibre angle during manufacturing, these materials can be easily tailored to mimic the anisotropy of the human mandible accurately (Schwartz-Dabney and Dechow, 2003). In a previous work (De Santis et al., 2004) it has been shown that by orienting glass fibres at $0^\circ/90^\circ$ and at $\pm 45^\circ$, Young's modulus of composites is 25 GPa and 7 GPa, respectively. Therefore, fibre orientation provides a very powerful tool to reproduce the anisotropy of the mandible cortical bone (Schwartz-Dabney and Dechow, 2003; De Santis et al., 2007).

Benefiting from this approach, the experimental model (De Santis et al., 2004, 2005) appeared useful to evaluate different materials and design of full-arch implant supported prostheses and their fit.

The amount of prosthetic misfit that peri-implant bone can tolerate without adverse complication is still unknown (Michalakis et al., 2003), and consequently it seems prudent to keep it as low as possible. Traditionally, prosthetic frameworks were manufactured with a cast gold technique that is inevitably exposed to distortion and consequent misfit; shrinkage of the impression material, thermal deformation of the mould, shrinkage of the metal due to the liquid–solid phase transformation and shrinkage of the solid metal during cooling (Michalakis et al., 2003) are all factors that might jeopardise the perfect fit of cast frameworks especially in case of long spans. Nowadays, frameworks obtained from industrial CAD/CAM processes are more frequently used to obtain titanium frameworks. In particular the Procera “All in one” technique is a very well established one with a 5-year follow-up study for the completely edentulous case, showing satisfactory clinical results (Ortorp et al., 2003; Ortorp and Jemt, 2004).

Similarly the present study confirmed a better fit for Procera titanium frameworks compared to cast gold ones, thus highlighting that CAD/CAM is effective in avoiding the formation of bone residual stresses that were instead observed with cast full-arch restorations. The strain gauges used to monitor local strains along the composite mandibles did not detect any strain at any of the five different titanium bridges once they were fitted on the implants, thus supporting passive fit reproducibility and avoidance of fit-related problems. It is worth noting that the present study compares the bone deformation caused by prosthetic misfit, also showing the possibility of evaluating the arising stress levels with precision of magnitude and location (Ortorp et al., 2003; Torsello et al., 2008). This is likely to be a much more relevant approach to establish a misfit threshold that might be considered harmful for the peri-implant bone. For instance, in the case of cast gold, it is possible that the strain generated from unavoidable technical-related misfit is of limited clinical harm. This would explain the positive clinical outcome associated with the technique and its still current use.

With regard to mandible deformation during the simulated action of pterygoid muscles, the effect of one- and two-piece Procera prosthesis design has been tested. Due to the activity of these muscles during mouth opening and closing the arch width decreases its amplitude (Chen et al., 2000; Murray et al., 1999; Phanachet et al., 2001) and, consequently, considering the average direction of the muscle lines of action, a lateral load between 10 N and 20 N is expected. Although these loads are very small (Koolstra and van Eijden et al., 1995; Murray et al., 1999) the resulting mandible deformations are still important since the deformation induced by the pterygoid muscles is also present during speech (Peck et al., 2000). Moreover the intensity of the pterygoid muscles is amplified by the mandibular length, that is the distance between the condyle and the mandibular arch (Chen et al., 2000).

Composite mandibles used in this study showed a stiffness of 35 N/mm, suitable to replicate the stiffness of edentulous mandibles (De Santis et al., 2004, 2005). Instead, when loading mandibles rehabilitated with one- and two-piece full-arch Procera prostheses, the stiffness of the system was modified by the prosthetic design (Fig. 4a and b).

In particular, one-piece prostheses in group A increased mandible stiffness by about 20% (Fig. 4a) probably ascribed to the stiffness of the one-piece titanium bridge. Although this increase of the stiffness is relatively small, it drastically perturbs stress distribution at specific sites of the mandible. In the premolar region, a remarkable stress-shielding effect (Apicella et al., 1998; Kennady et al., 1989) was observed (Fig. 4b). On the vestibular side the strain reduction for group A approximated 30% compared to the control group whilst, on the lingual side of the incisal region, higher strain reduction (about 50%) was recorded (Fig. 4b), also suggesting a stress-shielding in these areas due to the prosthesis. Therefore the investigated implants in conjunction with the full-arch restoration (group A), in the loading condition tested, acted as a force by-pass at the symphysis, as suggested by Apicella et al. (1998) and De Santis et al. (2005).

Conversely, when considering mandibles restored with Procera bridges, the stiffness of two-piece design in group B was significantly lower compared to one-piece design in group A (Fig. 4a), thus suggesting that both manufacturing technique and prosthesis configuration are important to determine changes in mandible biomechanics. Moreover the strain behaviour in incisal region in the group B resembled the control group (Fig. 4b) and no stress shielding effect was observed, thus supporting the importance of bridge configuration in determining biomechanical characteristics of the rehabilitated mandible. Stress distribution into mandibular bone rehabilitated with a one-piece full-arch bridge is more perturbed than that observed with a two-piece design. The effect of a specific rehabilitation on the implant–bone interface is of great concern, since both stress concentration and stress shielding might affect implant stability through bone necrosis and atrophy, respectively. For instance the association of one-piece prosthesis supported by implants placed posterior to the mental foramen resulted in a posterior implants rate of 40%, inexplicably higher than their anterior counterparts (Miyamoto et al., 2003). Accordingly, it was suggested that avoiding detrimental stress states to posterior implants due to mandible deformation requires particular attention in the number of planned posterior implants as well as in the necessity to split the prosthesis.

Finally, some limitations of the proposed model need to be discussed. The models reproduced the mechanical anisotropy of human mandibles, but they did not consider mechanical properties variation according to the site. In other words, the thickness of the composite shell and fibre density was uniform along the whole mandible. However, through the composite materials technology it could also be possible to vary the material stiffness according to the site.

In conclusion, precise measuring through strain gauges employed in our composite mandible model can be very useful to calibrate theoretical models developed to analyse the stress distribution around implants. In this scenario the study of the biomechanical effects of implants and arch prostheses can be useful to optimise the design of the prosthetic framework, thus improving treatment outcomes.

Conflict of interest statement

None of the authors have any conflict of interest with regard to this study.

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