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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 ±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 (±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.