

In Vivo Registration of Force Development with Ceramic and Acrylic Resin Occlusal Materials on Implant-Supported Protheses

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Purpose: It has been hypothesized that the shock generation on implant-supported protheses during chewing should generate higher implant loads if the veneering material is porcelain rather than acrylic resin. **Materials and Methods:** The present study uses strain-gauged abutments to measure the force transferred to the implant after a shock has been applied. This was measured in vitro and in vivo in 5 patients. **Results:** The different occlusal materials did not lead to different forces generated to the implants of the patients. **Discussion:** From a practical point of view, the choice of occlusal material has no bearing per se on force generation to the implants. **Conclusions:** The present study demonstrated that there: (a) is a difference in resilience between acrylic resin and ceramic veneering materials, but (b) this difference is only measurable in in vitro where the force is generated by a shock only and the implant is rigidly anchored. (INT J ORAL MAXILLOFAC IMPLANTS 2002;17:17-23)

Key words: biomechanics, dampening, resilience, strain-gauged abutment, veneering material

It has been hypothesized that the shock implant-supported protheses generate during chewing should generate higher implant loads when the veneering material is porcelain rather than acrylic resin.¹ There are, however, 2 theoretical factors making this hypothesis disputable: (1) the occlusal speed and jaw mass may not be sufficient to introduce any significant shock generation, and (2) the difference in elasticity between different occlusal materials may be overridden by the flexibility of the implant in the bone.

The force generated in mastication is developed by muscle actions. There is also a possibility that closure of the jaws may add an impulse force at the point of first contact. This impulse is proportional to the speed and the mass of the mandible and the

force developed is dependent on the resilience of the tooth system. Such resilience is found in the periodontal ligament of any antagonist natural tooth and flexibility of the implant bone anchorage, as well as that of the implant and prosthesis complex. From a force-transmission perspective, the implant-supported dentition could be seen as a series of springs attached to one another, each with different stiffness related to the mechanical properties of the structure components (Fig 1a). To ascertain whether such an impulse could contribute to and whether it is dependent on the veneering material, the force generated between the teeth and the rise time of that force are significant parameters to study. The masticatory force encompasses 1 high-frequency part (HF), ie, the shock at the initial contact, and 1 low-frequency part (LF), which is the grinding motion. From an engineering aspect,² the HF component will be more affected by changes in the total spring-system than the LF component.

Although there is published research related to the difference in stiffness of different materials,³⁻⁶ in vitro measurement of shock forces⁷ and in vivo occlusal contact measurement,^{8,9} no in vivo chewing force registration has been reported demonstrating such an effect.¹⁰ The question addressed in this

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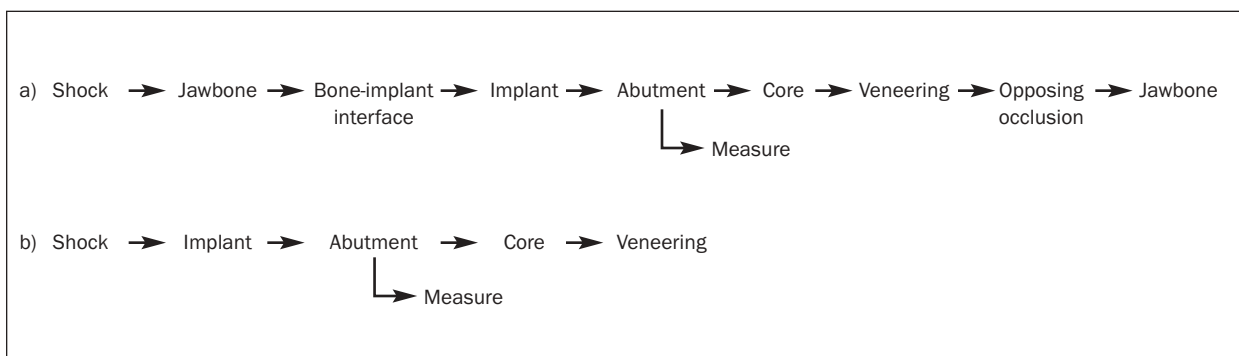


Fig 1 Structural components through which the shock will travel in vivo (a) and in vitro (b).

Table 1 Patient Data				
Patient	Age (y)	Edentulous region	Tested implant	Antagonist
1	59	Partial maxillary and mandibular	Maxillary right second premolar	MC prosthesis on implants 44 and 45
2	65	Complete maxillary and mandibular	Mandibular right second premolar	MC crown on implant 14
3	34	Maxillary left second premolar	Maxillary left second premolar	Natural tooth
4	38	Maxillary right second premolar	Maxillary right second premolar	Natural tooth
5	39	Maxillary left first premolar	Maxillary left first premolar	Natural tooth

study is whether the HF component is at all affected by the change in veneering material, and, if so, whether this component is significant enough compared to the LF component to cause an overall difference in vivo.

Recently a technique for in vivo registration of implant forces and moments has been developed.¹¹ In in vivo measurements on patients, the total force generated at various clenching and chewing situations was registered. In in vitro measurements a “worst-case” was studied (Fig 1b), ie, where the veneering materials were mounted in a stiff and nonresilient rig and with the only force applied being generated by a falling weight. The aim of the present study was to investigate whether this force transmission in vivo was influenced by different veneering materials, ie, acrylic resin or porcelain.

MATERIALS AND METHODS

Five patients previously treated with Brånemark implants in the posterior region comprise the test

individuals (Table 1). For each patient, new prostheses were fabricated for the restoration. The premolar position was arbitrarily selected to be the test site. The occlusal force is larger in more posterior positions, while the occlusal speed is larger in more anterior positions, because of the lever mechanism of the mandibular joint. The premolar position will thus give a balanced mix of muscle and shock forces.

Two single tooth crowns were fabricated for each patient; 1 with porcelain and 1 with acrylic resin (Figs 2a to 2c). The acrylic resin crown was made of polymethyl methacrylate resin (Meliodent, Heraeus Kulzer, Wehrheim, Germany) on a standard temporary plastic coping (Brånemark System, Nobel Biocare, Göteborg, Sweden) and the ceramic (Vintage, Shofu, Kyoto, Japan) crown was based on ceramic gold alloy (Estheticor Ideal H, Cendres & Metaux, Bel-Bienne, Switzerland) cast on a standard 4-mm gold cylinder (Brånemark System, Nobel Biocare, Göteborg, Sweden). This setup is believed to maximize the content of acrylic resin and porcelain in the 2 crowns. The acrylic resin crown was made by duplicating the ceramic one using the Ticonium



Fig 2a (Left) In vivo picture (patient 4). The specially designed test prosthesis with the measurement strain gauged abutment in place. Note the supra occlusion of the test prosthesis. The strain gauge is indicated by an arrow.

Fig 2b (Above) Radiograph of the ceramic test prosthesis with the strain gauged abutment in place (patient 4).



Fig 2c Radiograph of the acrylic resin test prosthesis (patient 4). Note that the acrylic resin part of the prosthesis is not visible on radiograph.

duplicating flasks technique with agar-agar hydrocolloid duplicating material,¹² thus trying to make the 2 crowns as similar as possible. The maximum deviation was estimated to be of the order of 10 μ m. The single tooth prostheses were attached to standard 5.5-mm abutments equipped with 3 strain gauges ad modum Glantz and associates.¹¹

To ensure that the occlusal forces were transmitted through the test prostheses, these prostheses were adjusted to supra- and centric occlusion, so they would establish the first and main contact in occlusion/chewing. The supraocclusion was 0.5 to 1 mm in 3 cases; in 1 case it was 2 mm. The opposing dentition is shown in Table 1. The patients were asked to apply 1 rapid biting motion (Fig 3a) on the test-implant prosthesis, as well as several rapid chewing motions (Fig 3b). The applied muscle forces and shocks were registered as strains in the measurement abutments and the corresponding axial forces were calculated.^{9,11}

The research protocol was also applied to an in vitro model (simulating stiff implant anchorage) in which the implant was placed in a steel plate and the jaw impact was generated by a weight of 164 g falling from a standardized height (Fig 4).

The measured data were analyzed in 2 different ways:

1. *Peak force.* The peak forces reached in maximum occlusion were compared between the acrylic resin and ceramic restorations for each patient. The *peak-force ratio* is defined as the ratio between 2 peak forces (acrylic/ceramic). Hence a peak-force ratio of 70% means that the acrylic

resin peak force is only 70% of that of the ceramic veneering (Fig 5).

2. *Rise-time delay.* The time from application of the force until the peak force was reached was measured (Fig 5).⁷

RESULTS

The in vivo measurements were somewhat compromised by a 50-Hz noise pattern, which was believed to come from the external power supply. This noise-pattern was filtered out after the measurements. Apart from this, the measurement system was satisfactory from a technical point of view.

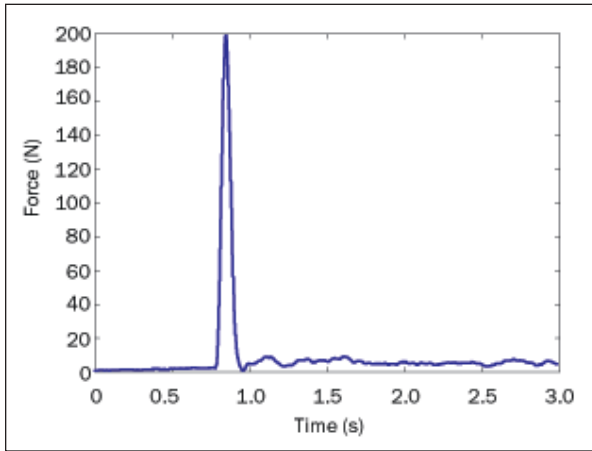


Fig 3a One rapid bite measurement, patient 1.

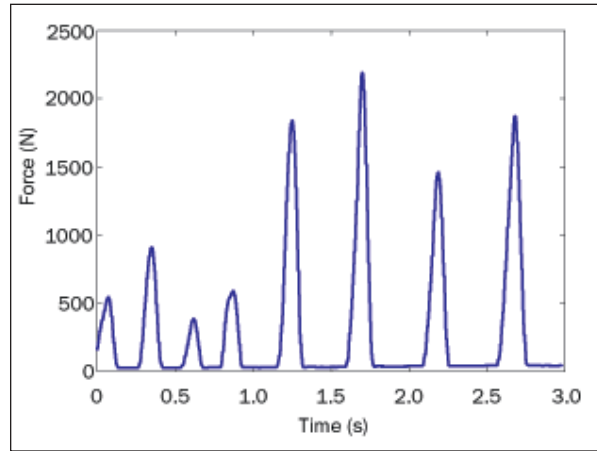


Fig 3b Several chewing motions, patient 4. Note the large variation within the measurement.



Fig 4 In vitro model. The hammer falls from a standardized height onto the prosthesis mounted on a strain gauged abutment.

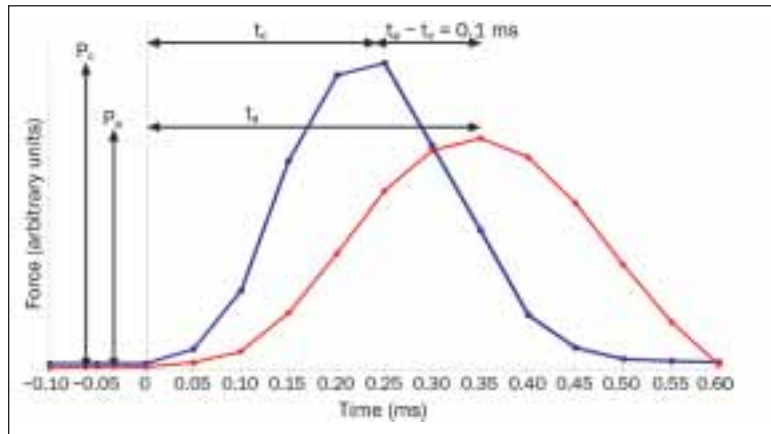


Fig 5 In vitro measurement. This measurement illustrates the 0.1-ms rise-time difference between an acrylic resin (*red*) and ceramic (*blue*) veneering in vitro. The figure also defines rise time and peak-force ratio. The rise time is defined as the time from the start of the shock until the peak force, indicated by t . This delay in rise time is indicated by $t_a - t_c$, where a indicates acrylic resin veneering and c ceramic. The peak forces are indicated by P and the peak-force ratio is defined as P_a/P_c .

The in vitro measurements showed a good signal-to-noise ratio. In addition, the measurement showed good reproducibility (Fig 6), suggesting a sound measurement method.

Peak-Force Comparison

The average and standard deviation of the in vivo applied peak forces within each patient and type of restoration are shown in Table 2. The in vivo measurements showed a large variation in peak applied force and no statistical significant difference was found between the different veneering materials. The peak force ratios varied between 20% and 130%.

The same information can be found in Table 3 for the in vitro measurement. The measurements were highly repeatable and showed a much higher peak force for the ceramic veneering material. The peak-force ratio was in the range of 10% to 35%; ie, the acrylic resin peak forces were 10% to 35% of the corresponding ceramic (Figs 5 and 6).

Rise-Time Comparison

The in vivo rise times were all of the order of 0.1 seconds, irrespective of veneering materials. No statistically significant difference between the groups could be seen. This is demonstrated in Fig 7, where 2 measurements from the same patient demonstrat-

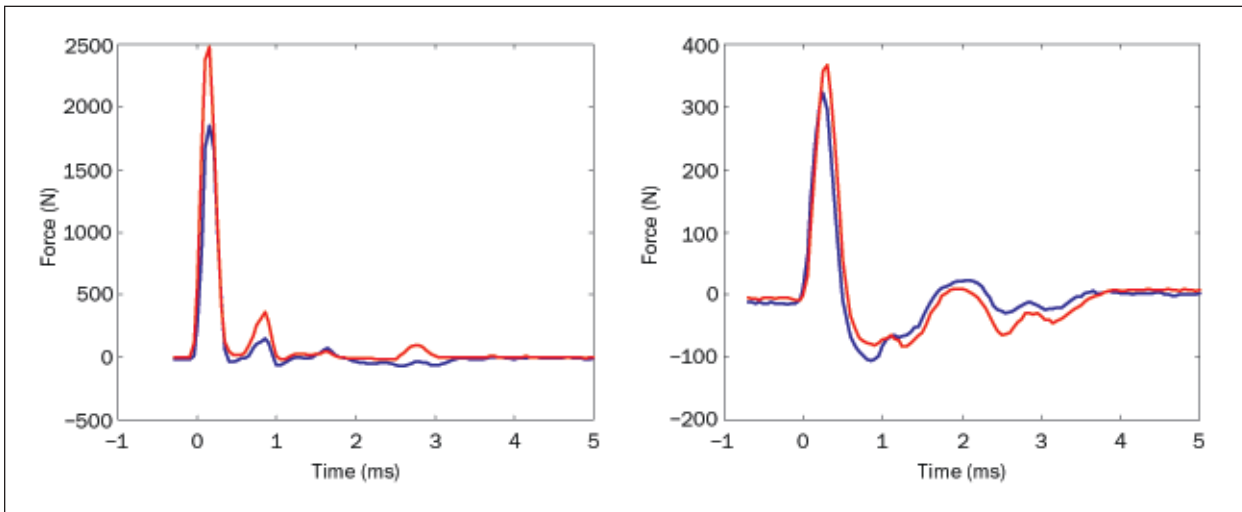


Fig 6 Reproducibility of in vitro measurements. The figure shows 2 ceramic (*left*) and acrylic resin (*right*) in vitro measurements using the same setup. Note the reproducibility between the measurements.

Table 2 Peak Forces of In Vivo Measurements						
Patient	Material	N	Peak force (N)		Rise time (s)	
			Average	SD	Average	SD
1	Acrylic resin	3	198	16	0.02	0.04
	Ceramic	2	195	11	0.02	0.07
2	Acrylic resin	2	154	42	0.025	0.002
	Ceramic	1	63		0.032	
3	Acrylic resin	7	450	315	0.060	0.039
	Ceramic	1	96		0.025	
4	Acrylic resin	9	2055	820	0.113	0.014
	Ceramic	24	1039	601	0.111	0.036
5	Acrylic resin	14	995	589	0.069	0.033
	Ceramic	9	1280	466	0.081	0.031

Note the high standard deviations of the measurements (compare with Fig 3b). There are no statistically significant differences between the veneering materials in any patient. N = number of peaks for each patient and veneering material. SD = standard deviation.

Table 3 In Vitro Data					
Patient	Rise time (ms)		Forces (N)		Peak force ratio (%)
	Ceramic	Acrylic resin	Ceramic	Acrylic resin	
1	0.18	0.95	310	110	35
2	0.30	0.33	3025	315	10
3	0.20	0.30	1197	403	34
4	0.28	0.33	3668	530	15
5	0.20	0.33	2164	344	16

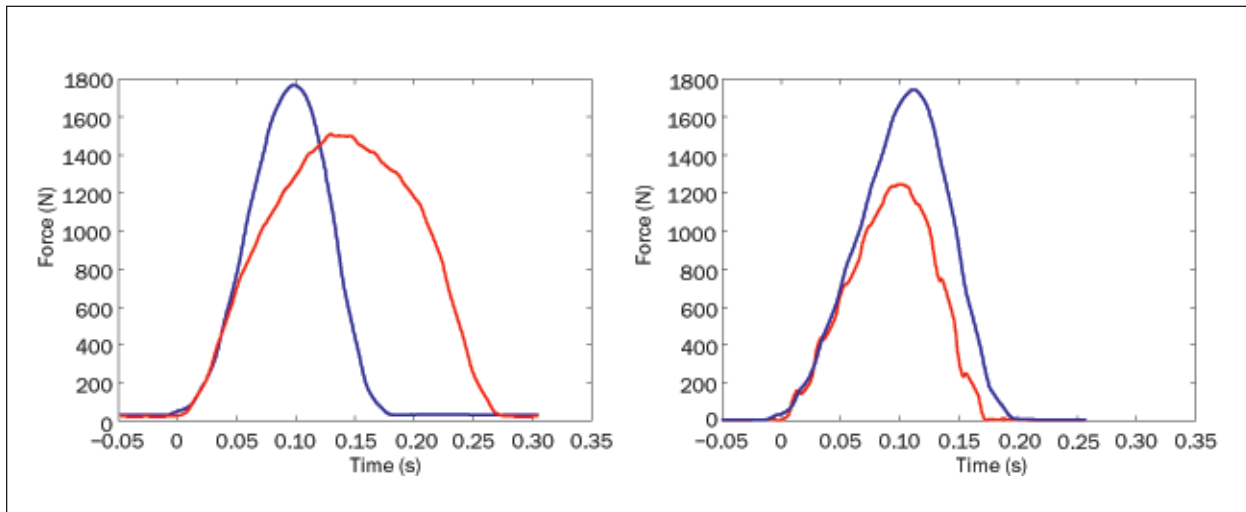


Fig 7 Dampening (*left*) and no dampening (*right*) within the same patient (patient 4).

ing faster rise times for ceramic (Fig 7a) and acrylic resin (Fig 7b) are shown.

In the *in vitro* study, the rise times were of the order of 200 μ s for the porcelain and 300 μ s for the acrylic resin veneering material, respectively (Table 3). Thus, a delay of the peak force of approximately 100 μ s was demonstrated for the acrylic resin material (Fig 5).

DISCUSSION

The most important finding from the study was that the different occlusal materials did not lead to different forces generated to the implants in the patients. Therefore, from a practical point of view, the choice of occlusal material apparently has no bearing *per se* on force generation to the implants.

In the *in vivo* measurements, the differences in generated forces within a number of chewing cycles were far greater than any differences related to different occlusal surfaces (Fig 7). The rise times of the *in vivo* measurements were all of the order of 0.1 seconds, which is comparable to other reports⁹ describing normal chewing cycles. There was no statistical difference in rise time between the groups.

The explanation for the vast differences within the same patient pertains to the result of differences in the exact occlusal contact between the measurements, as well as the relaxation of the patient, *ie*, his or her ability to apply the desired rapid shock. To ensure that the occlusal forces were generated at the

test prostheses they were put in supraocclusion, which *per se* leads to a nonfamiliar occlusal situation for the patient. This, in turn, may have contributed to the variation in force generation.

The rise times between the *in vivo* and *in vitro* measurements are dramatically different: 0.1 seconds *in vivo* compared to 300 μ s *in vitro*, hence a factor of at least 300 μ s difference. This means that the *in vitro* shock consisted of much higher frequency components than the *in vivo* shock. This is reasonable, since a falling weight (the impulse) was the only force generator and there was no force simulating the muscle action. Further, bone resilience and resilience of the tooth antagonist were omitted.

The differences between the *in vivo* and *in vitro* model are mainly resilience of the jaws and the force applied by muscles as compared to a falling weight. Both of these factors exert a heavy impact on any extrapolation of the *in vitro* data to clinical reality. Concluding from the results of the *in vitro* and *in vivo* measurements, the main finding of the present study was that even if there is a difference in resilience between acrylic resin and ceramic veneering materials, there apparently is no clinical relevance. It is hypothesized that this difference in dampening is only measurable for peaks consisting of such high frequencies, *ie*, fast movements, that they cannot be generated by the human jaws—at least not as a conscious act of will. This hypothesis is supported by the fact that most natural mechanical systems are a more effective damper for higher than for lower frequencies.² It is also hypothesized that the resilience of the masticatory system is overriding

the stiffness difference of the various materials.

Gracis and coworkers⁷ measured dampening in vitro, using even faster shocks than used in the present study. Although the materials and methods used in this study were somewhat different than those of the present, the general results are comparable on an order-of-magnitude scale. In the study by Gracis and coworkers, an applied weight of 1.06 g caused an impulse with a rise time of the order of 15 μ s. This impulse resulted in a peak force of 140N and 60N for the ceramic and acrylic resin groups, respectively. Hence, the dampening was of the order of 45%. The corresponding figure for the present study, as seen in Table 3, is in the range of 10% to 35%. Hence, 2 factors can be noted: The difference in dampening between the acrylic resin and ceramic veneers was amplified in the present study and the variance between samples/patients was increased. The amplified difference is believed to be the result of the heavier applied weight in the present study, 164 g as compared to 1.06 g. The increased variance is believed to be the result of the individual crowns. With the same crown, the difference between 2 measurements was very low (Fig 6). From both studies, it can be concluded that in an in vitro environment it can be demonstrated that an acrylic resin veneer damps the applied shocks more than a ceramic veneer. However, both in vitro studies had rise times of at least a factor 300 μ s faster than what is normally seen in vivo.

Skalak¹ hypothesized that a soft layer on a prosthesis, such as plastic, could reduce the peak forces generated by the shocks applied during mastication. He also suggested that an acrylic resin veneering might have this effect. The conclusion of the present study is that the resilience of an acrylic resin veneer is not sufficient to cause any significant change in the force transmission through the prosthesis as compared to a ceramic veneer in vivo. To demonstrate such an effect, an in vitro model with faster shocks and/or less resilience in the supporting structures than in vivo was needed.

CONCLUSIONS

From the present study it was concluded that the different occlusal materials used did not lead to dif-

ferent forces generated to the implants in the patients. Therefore, from a practical point of view, the choice of occlusal material apparently has no bearing per se on force generation to the implants.

Further, the present study demonstrated that there: (a) was a difference in resilience between acrylic resin and ceramic veneering materials, but (b) this difference was only measurable in in vitro models, where the force is generated by a shock only and the implant is rigidly anchored.

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