

Finite element simulation of fixed dental prostheses made from PMMA —Part II: Material modeling and nonlinear finite element analysis

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Material characteristics can change significantly with increasing chewing velocity. As these *in-vitro* examinations are very time-consuming and cost-intensive, the application of finite element analysis (FEA) offers a suitable alternative for predicting the material behavior of complex specimen geometries under clinically relevant loads. Although FEA is applied within numerous dental investigations, there are only few studies available in which a nonlinear FEA is validated with real experiments. Therefore, the aim of the present study was to predict the mechanical behavior of a clinically close three-unit temporary bridge composed of polymethyl methacrylate (PMMA) in the left upper jaw with nonlinear FEA and to verify the prediction through validation experiments. In conclusion, simplifying assumptions of linear elastic material properties for polymeric materials should be avoided in FEA studies, because rate dependencies, stress relaxation and plastic flow are not considered. Additionally, precise preliminary investigations for material characterization are necessary.

Keywords: Finite element analysis, Materials testing, Dental materials, Materials science, Prosthodontics

INTRODUCTION

Test methods in dental biomaterial science are still very traditional, relying on real experiments, which poses a problem because *in vitro* test series are usually very time-consuming and cost-intensive and require large amounts of resources. In addition to the breaking load, which is directly dependent on the specimen geometry, the bending strength is often used, making it possible to make a statement about the strength of a material¹⁾. A disadvantage of using only this parameter, however, is that no statements can be drawn about the stability of a complex specimen geometry, such as a bridge. Therefore, to optimize the stability and to predict the probability of failure, a method with which even complex geometries can be analyzed before manufacturing, is necessary. This possibility is offered by finite element analysis (FEA)²⁾.

Thus, in analogy to the 3R principle (Replace, Reduce, Refine) by Russel and Burch, which is used in animal testing, it would be a decisive step forward if FEA, which is a well-established method in engineering fields, is applied to aspects regarding implantology. Unfortunately, it is yet to be implemented to a large extent in biomaterial science^{3,4)}. FEA is a numerical approximation method, with the help of which stress distribution, among other things, in a component with complex geometry under load can be calculated. Nonetheless, a problem arises in a clinical setup, where there are different hard-to-assess boundary conditions for tissues and their properties, requiring every FEA model to be verified. Therefore, our work aimed to analyze this method meticulously and

to investigate if the application of FEA, in principle, is a feasible approach for dental biomaterial science. To compare simulation data to a real model, a traditional model was needed for crosschecking. However, because the technology for the application of FEA is available, we used a model of a three-unit fixed partial denture (FPD) composed of PMMA, which is used for similar approaches to dental biomaterials, to analyze whether time-consuming and costly laboratory investigations can be minimized using this model calculation. Because the entire project is very complex, Part II of this paper focuses solely on the FEA.

The use of correct material properties is of fundamental importance in FEA calculations. In the case of complex material behavior, material characterization may require extensive test series. For simplification purposes, however, a purely linear elastic material behavior is often assumed instead⁵⁻⁸⁾. On the other hand, however, simplifying assumptions can lead to a limited validity for the calculated results. For this reason, the material properties of the material used (Telio CAD) were first examined in a previous work⁹⁾. In this study it was shown that the material behavior of the used PMMA is strongly nonlinear, because the material exhibits stress relaxation, plastic deformation and a rate-dependent elastic modulus. Within the study, linear elastic simulations on three-point bending specimens were also performed. The simulation results differed largely from the experimental results, suggesting that a more complex material model is needed to achieve accurate simulation results. Furthermore, these observations indicate that the material cannot be simulated realistically with a linear elastic model. Because FEA is

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a very complex investigation method, the already well-investigated polymethyl methacrylate (PMMA), which is typically applied in provisional restorations¹⁰, was used, primarily to develop a model rather than to investigate the material itself. Otherwise, too many surrounding conditions can lead to altered results that cannot be directly attributed to the material properties and that can therefore be misinterpreted. The main question in the development and application of a new dental biomaterial is whether this material can withstand the chewing loads from a patient over a long period of time. However, pure statements by various manufacturers about minimum thickness are useless if the stress state in the material is extraordinarily complex and if the behavior of the material is influenced by stress triaxiality. In this case, even when the specified material thickness is used, fractures can occur. The prediction of a possible fracture can be performed with the help of FEA. In the future, integration into intraoral scanner software or CAD software in the dental laboratory may be feasible. This would make it possible to determine, with the help of FEA, whether a material would be able to withstand masticatory loads, even in the complex case of a stress state before any dental prosthesis is created.

Although FEA is applied within numerous dental investigations, there are only few studies available in this field in which FEA results are compared with real validation experiments^{11,12}. Furthermore, FEA is only used in academic research fields and not in daily dental practice under clinically closed conditions. Therefore, the aim of the present study was to predict the mechanical behavior of a clinically close three-unit temporary bridge composed of PMMA in the left upper jaw by using nonlinear FEA and to verify the prediction through validation experiments.

MATERIALS AND METHODS

Three-point bending simulation and material modeling

In preliminary investigations, the cross-linked PMMA material with percentage weight 99.5%, <1% pigments with no further fillers (Telio CAD, shade A3, Ivoclar Vivadent, Schaan, Lichtenstein)^{13–15} was tested *via* three-point bending (3PB) tests and dynamic mechanical technical analysis (DMTA)⁹. A 3PB simulation model was set up to determine the material parameters and validate the material cards resulting from different experiments that were performed in the past⁹.

The computational study was conducted *via* FEA using pre- and postprocessor LS-PrePost and the implicit finite element solver LS-DYNA (LSTC, Livermore, CA, USA). As shown in Fig. 1, the 3PB test specimens were modeled using solid elements with a mesh size of 0.2 mm (20,000 solid elements). Both the bending fin and bearing were considered as rigid shells composed of steel (elastic modulus $E=210$ GPa; Poisson's ratio $\nu=0.3$; density $\rho=7,850$ kg/m³) and were meshed with a mesh size of 0.1 mm and a radius of 1.0 mm (3,720 shell elements). The contact between the specimen and fins was modeled using a two-sided automatic surface-

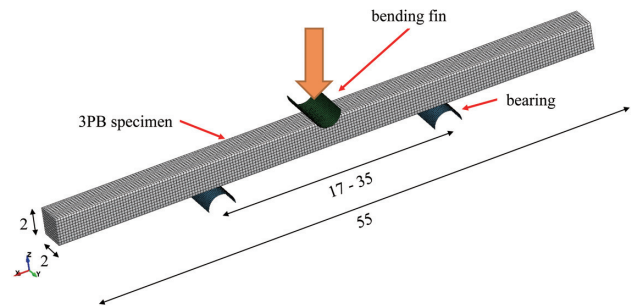


Fig. 1 Modeling of 3PB specimens (length values in [mm]).

to-surface mortar contact. The specimen geometry, the distance of the bearing and the loading velocity of the bending fin were input to the model according to the 3PB test setups performed by Schmidt *et al.*⁹.

To simulate the material behavior of Telio CAD, a material card based on the semi-analytical model for polymers (SAMP) was created¹⁶. The special features of this material model are that it can describe both viscoelastic effects (in this case, especially, stress relaxation and the rate-dependency of elastic modulus) and viscoplastic behavior. Although not considered in this study, the SAMP also supports stress-state-dependent plastic flow.

The rate-dependent elastic modulus was modeled in the simulations using

$$E(\dot{\epsilon}) = [E_{qs} - E_{dyn}] \exp(\alpha_E \dot{\epsilon}^{\beta_E}) + E_{dyn} \quad (\text{Formula 1})$$

as an adaptation of the relationship proposed by Schmidt *et al.*⁹. Here, E_{qs} is the elastic modulus from the quasistatic 3PB test, E_{dyn} is the elastic modulus at the maximum loading rate, and α_E and β_E are the fit parameters. For the yield condition, the von Mises yield criterion was used with the tabulated yield stress over the effective plastic strain. However, given that the plastic strain cannot be accurately estimated from the 3PB tests, the plastic behavior was determined *via* reverse engineering. Therefore, the curves of yield stress σ_s over the effective plastic strain ϵ_p were modeled using Schmachtenberg's hardening law¹⁷

$$\sigma_s = \sigma_y + c_1 \epsilon_p \frac{1}{1 - c_2 \epsilon_p} \quad (\text{Formula 2})$$

with the yield stress σ_y , and the two hardening parameters c_1 and c_2 . The parameters were optimized to fit the 3PB test results of Schmidt *et al.*⁹. Furthermore, to accompany stress relaxation, the viscoelastic decay coefficient was determined *via* a relaxation test.

Relaxation test

For the relaxation test (RT), a Telio CAD specimen of dimensions 20×2×55 mm was clamped into a servo electric universal testing machine, Inspekt 5 (Hegewald & Peschke, Nossen, Germany) with a clamping length of

15 mm. The specimen was loaded with a force rate of 10 N/s until a strain of 0.3% was reached. This deformation was maintained for 5 h. The decay coefficient was calculated from the reciprocal value of the relaxation time, that is, the time after which 63.2% of the applied overstress has decayed.

Simulation of dental prostheses

Based on an idealized chamfer preparation (preparation taper 6°; stump height 6 mm; anatomical circular and occlusal substance removal 1 mm) a three-unit FPD for teeth FDI 25 and 27 with missing tooth 26, as patient equivalent was created using computer-aided design (CAD; millhouse, Hofheim-Wallau, Germany). Afterwards, the FPD was produced *via* computer-aided manufacturing (CAM)¹⁸⁾. The FPD was discretized with roughly 130,000 solid elements. Further, tetrahedron elements were used because these shapes deliver the best spatial adaptability at the boundary surfaces of the geometry. In this study, linear tetrahedrons and quadratic tetrahedrons were used for the simulation of dental prostheses. Further, their levels of performance were compared.

The linear tetrahedron element (T4) is spanned by four nodes in the corners of the tetrahedron. Here, each node has three degrees of translational freedom in the direction of the global coordinates. The interpolation of the node values is performed *via* the superimposition of linear form functions. In the simulations, a one-point constant stress tetrahedron with nodal pressure averaging was used.

A considerably higher result quality may technically be achieved *via* interpolation with quadratic form functions, which are used with solid elements, with center nodes on the element edges. For the quadratic elements (T10), six additional nodes are inserted in the middle of each element edge. Thus, the node values are interpolated through ten quadratic form functions. In general, the T10 elements are considered to be more accurate than the 4-noded tetrahedron elements. Unfortunately, this increase in accuracy leads to higher computational costs.

In the given simulation setup, the antagonist was modeled with a metal sphere, whose center coordinate had been determined in preliminary investigations [Part I: Experimental investigation under quasi-static loading and chewing velocities]. Within the simulation, the ball was modeled with about 2,900 fully integrated shell elements. The metal sphere was considered as a rigid body with the material properties of steel ($E=210$ GPa, $\nu=0.3$, $\rho=7,850$ kg/m³). Contact between the prosthesis and antagonist was achieved *via* automatic surface-to-surface mortar contact. The sphere was translated with a constant velocity that was determined using the velocity of the testing machine. In Fig. 2, the simulation setup, including the bridge and antagonist, is shown. In contrast to the component test from Part I of our study¹⁸⁾, the sphere was constrained in all its degrees of freedom except its movement in the load direction. The mounting of the bridge was achieved using

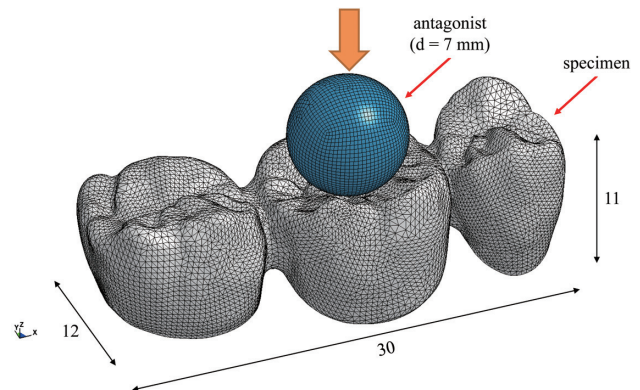


Fig. 2 Simulation setup of test specimen (length values in [mm]).

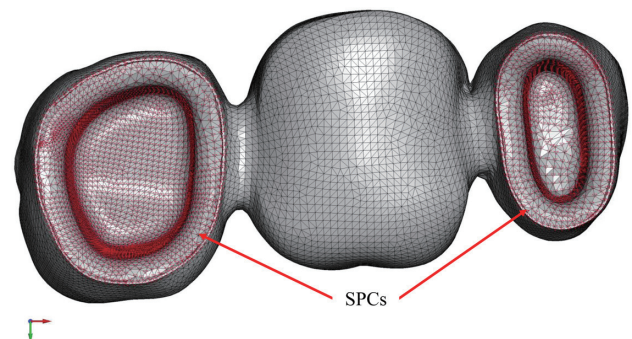


Fig. 3 Test specimen with marked SPCs.

single-point constraints (SPCs) that are constraint in all degrees of freedom, neglecting the possible influences of the sintered specimen holder and the adhesive bond from the setup in Part I of this study¹⁸⁾. The mounting of the bridge using SPCs is highlighted in Fig. 3. Analogous to the 3PB simulation, the model was set up and solved with pre- and postprocessor LS-PrePost and the implicit finite element solver of LS-DYNA (LSTC).

Statistical analysis

Forces at the different displacements obtained from the experimental and simulated data were compared statistically using Pearson's correlation tests and the intraclass correlation coefficient, additionally. Statistical analysis was performed using SPSS Statistics 25 (IBM, Armonk, NY, USA).

RESULTS

Relaxation test

As shown in Fig. 4, the calculated stress decayed from an initial maximum of 25 MPa to a value of 20 MPa. A relaxation time of 1,465 s can be determined from the measurements, corresponding to an overall decay coefficient of 6.83×10^{-4} 1/s.

Three-point bending simulation

In Fig. 5, the relationship between the elastic modulus and strain rate is shown with the fitted model following Formula 1 ($E_{qs}=2,594.19$ MPa, $E_{dyn}=4,723.99$ MPa, $\alpha_E=-1.3794$, $\beta_E=0.5792$) and the relationship determined by Schmidt *et al.*⁹⁾. The numerically determined curves of the yield stress over the effective plastic strain are shown in Fig. 6. The determined parameters for the

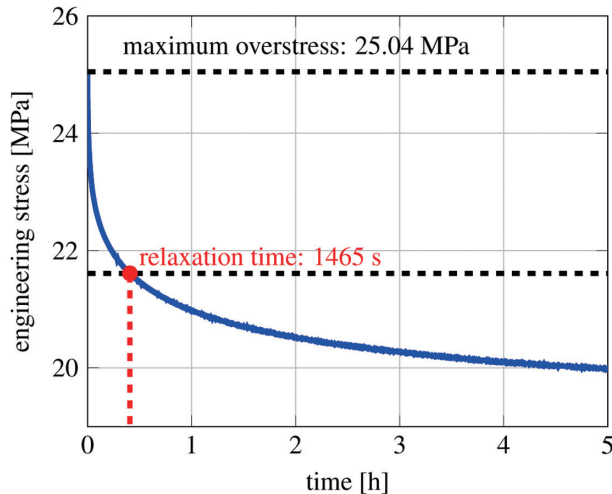


Fig. 4 Relaxation test result.

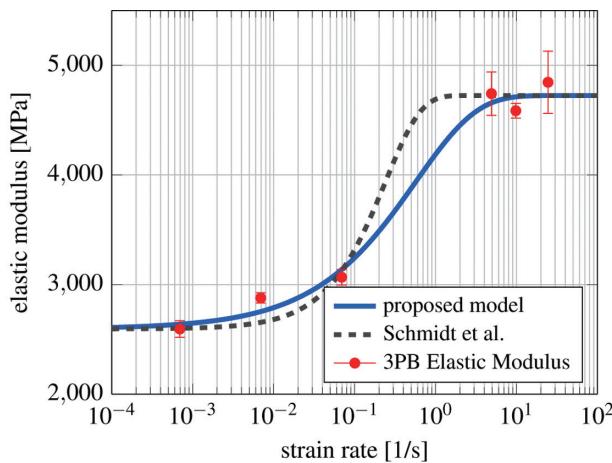


Fig. 5 Fitted relationship between elastic modulus and strain rate.

Schmachtenberg's hardening law are listed in Table 1.

The resulting force–displacement curves following the simulation of the 3PB specimens are shown in Figs. 7 and 8. As shown, the force–displacement curves can be reproduced in good agreement with the determined material model.

Finite element simulation of the component test

The force–displacement curves resulting from the simulations at loading velocities of 1.0 mm/min (0.0166 mm/s) and 130 mm/s (7,800 mm/min) are shown in Fig. 9 (Pearson's correlation tests 0.997 between both simulations and the experimental mean; intraclass correlation coefficient 0.999) and Fig. 10 (Pearson's correlation tests between T4 simulation and experimental mean=0.988, between T10 simulation and experimental mean=0.991; intraclass correlation coefficient 0.985), respectively. Furthermore, both diagrams show the mean curve of the test results from Part I of our study¹⁸⁾ with the associated standard deviation. The standard deviation of the mean curve obtained from the experiments is indicated by the blue markers. The red cross at the end of the experimental mean is intended to indicate the maximum and minimum forces and displacements at fracture measured during the experiments.

In Fig. 11, a comparison between the strain field obtained from the simulation with T4 elements and the DIC measurements from Part I¹⁸⁾ is visualized at 60 s

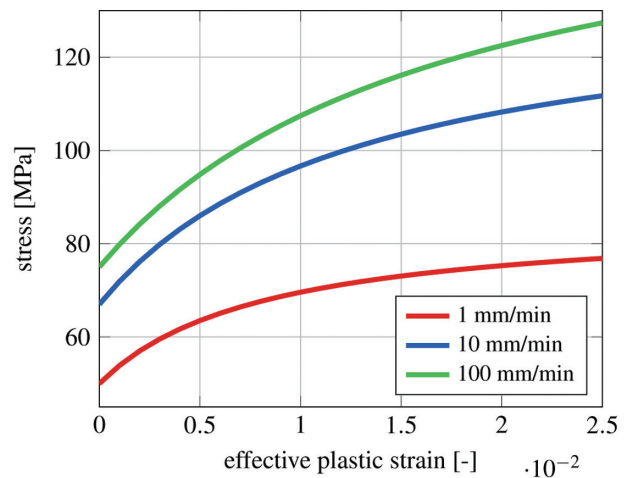


Fig. 6 Calculated curves of yield stress *vs.* effective plastic strain.

Table 1 Identified material parameters for Schmachtenberg's hardening law

| Test velocity [mm/min] | σ_y [MPa] | c_1 | c_2 |
|------------------------|------------------|---------|---------|
| 1 | 50.0 | 4,352.6 | -122.13 |
| 10 | 67.0 | 5,286.8 | -78.23 |
| 100 | 75.0 | 5,112.9 | -57.66 |

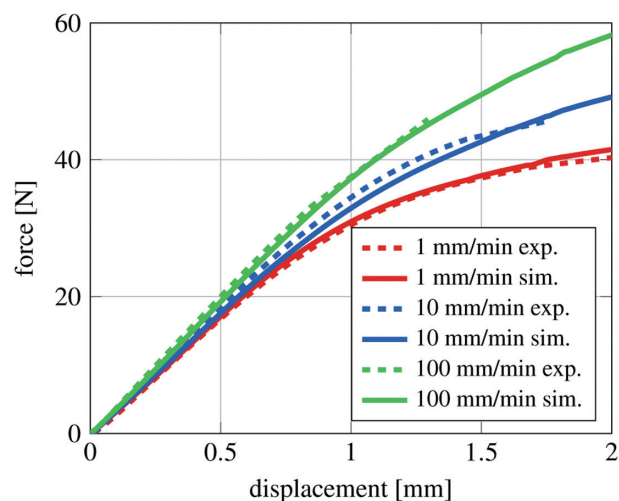


Fig. 7 Force–displacement curves for experimental and simulated 3PB specimens (low velocities).

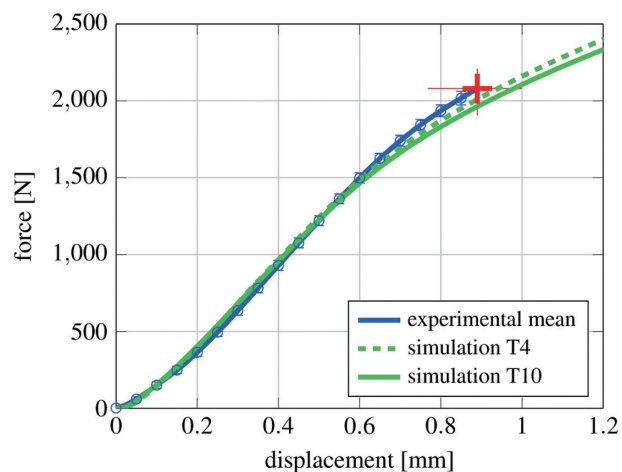


Fig. 9 Comparison between experimentally determined and simulated force–displacement curves of the specimen (1.0 mm/min).

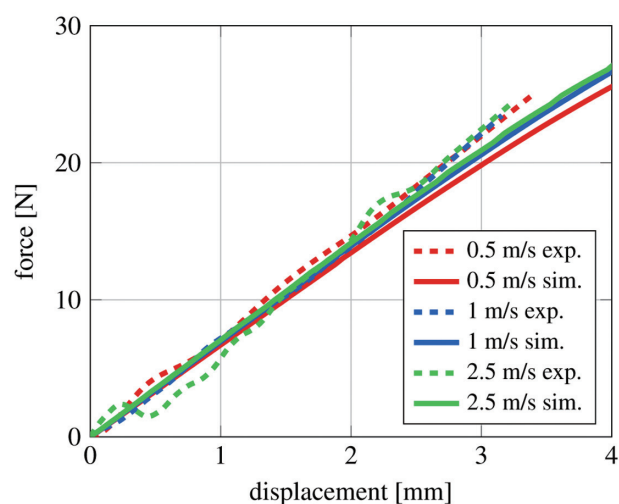


Fig. 8 Force–displacement curves for experimental and simulated 3PB specimens (high velocities).

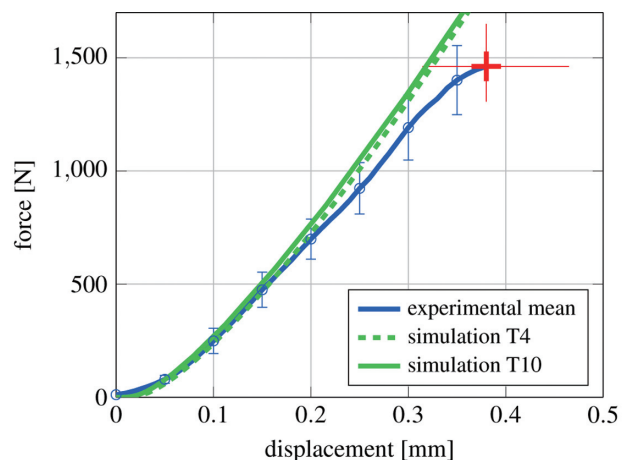


Fig. 10 Comparison between experimentally determined and simulated force–displacement curves of the specimen (130 mm/s).

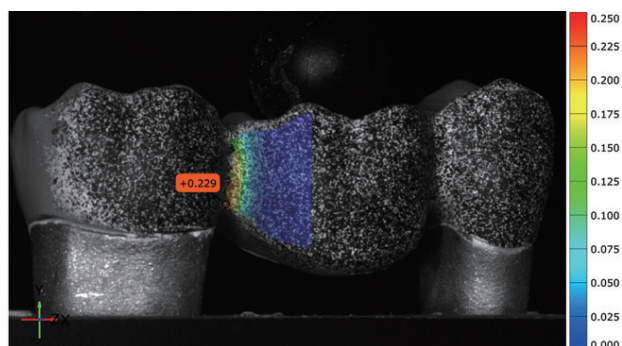
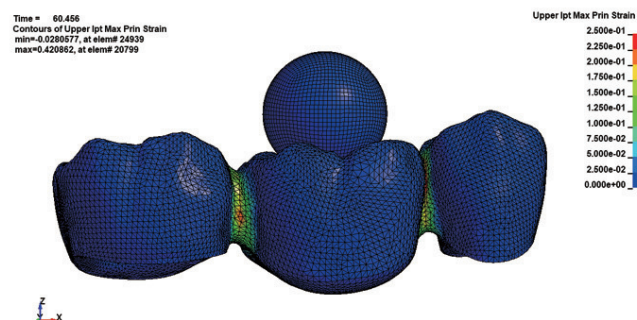


Fig. 11 Comparison of strain fields between simulation (left) and experimental setup (right).

of the test run time. As stated in Part I¹⁸⁾, the size of the facets used in DIC corresponds to a mesh size of approximately 0.4 mm, which is slightly larger than that in the FE simulation (about 0.35 mm element size in the area of the indentations). It should be noted that the legends in both the simulation and experiment are set to the same color values to enable a visual comparison.

DISCUSSION

Even if a detailed discussion of the SAMP material model would extend beyond the scope of this work, the results of this study indicate that the viscoelastic and viscoplastic properties of PMMA should not be neglected in dental biomaterial science. This mainly concerns much more complex materials, such as hybrid ceramics or CAD/CAM-composites with ceramic network structure or ceramic particles in PMMA. Therefore, the behavior of PMMA is of great importance as a fundamental component. In the following, we will discuss the most important points concerning the creation of the material card and the implications in dental material science.

As can be observed in Figs. 7 and 8, the force–displacement curves measured in the 3PB tests can be reproduced in good agreement with the experimentally determined SAMP. The initial slopes of the mean curves are captured with good accuracy at almost all machine traverse velocities, leading to the conclusion that the relationship from Formula 1 has at least validity in terms of explaining the observed phenomena. As shown in Fig. 5, the introduction of a second parameter to the model of Schmidt *et al.* leads to a much better accuracy for the relationship of elastic modulus over strain rate⁹⁾.

Given that stress relaxation could have a significant influence on the material behavior, the use of stress relaxation as accompaniment for FEA is expected to be advantageous. However, caution must be exercised, because the deformation history also influences the stress relaxation of the material. This means that, for example, in the case of a relaxation test, different loading velocities and displacements will likely lead to different relaxation times. This cannot be explicitly reproduced by the material model used here. However, this influence is assumed to be negligible for this given material, especially in the context of the good agreement between the simulation and experimental results in Fig. 7.

Regarding the relaxation test, it should be noted that the relaxation time was only qualitatively determined with one specimen in a tensile setup. A more intensive investigation of the relaxation behavior of the material was refrained from, mainly because the SAMP could not simulate such behavior anyhow. In particular, as other authors have already reported on the importance of relaxation experiments for dental biomaterial science¹⁹⁾, it might be useful to investigate relaxation behavior more extensively in future studies. It should also be mentioned that the relaxation test could have also been conducted in a 3PB setup. However, to qualitatively determine the relaxation time within this study, a tensile setup was

considered more pragmatic.

In this study, Schmachtenberg's hardening law was used to model plastic hardening, because the model is relatively simple and dependent on only three parameters. This selection is deemed reasonable, given that PMMA has been shown to exhibit strain hardening²⁰⁾. The Schmachtenberg's hardening law delivers satisfactory results for the simulation of the 3PB tests. Unfortunately, in the component test at 1 mm/min, the maximum force was slightly underestimated by the material model. This may be attributed to the selection of a von Mises yield criterion with stress-state-independent plastic flow. The von Mises yield criterion is mostly used to model metal plasticity, which in the general case is independent of the hydrostatic stress state and characterized by isochoric deformations. Polymers do not behave in this manner for the most part, resulting in a yield stress which is dependent on the hydrostatic stress-state within the material (*e.g.* yield under tension occurs under lower stress than yield under compression, which is also referred to as tension-compression-asymmetry). However, depending on the polymer, the usage of the von Mises criterion can be a sufficiently accurate approximation. As we aimed for the determination of a material model which could be determined from 3PB tests alone, only one stress state was tested, wherefore the von Mises yield criterion was selected. Considering the plastic deformations of the material under pressure might improve the simulation results of the prosthesis. However, this would strongly increase the experimental effort, because compression tests must then be performed and implemented, whereas, with the presented approach, the material could be modeled solely from 3PB tests.

Within the simulation setup of the prosthesis, some simplifications had to be made to minimize possible influences from factors other than the overall material behavior. In this study, the mounting of the bridges was performed with SPCs, because this could strongly decrease computation times, while achieving sufficiently accurate simulation results. Nonetheless, this might further cause deviations between FEA and experiment, especially given the overall stiffness of the model.

For the simulation of the low-velocity component test, as shown in Fig. 9, the measured force–displacement curve is in good agreement with the simulation model. Further, it can be observed from the simulation results that the influence of the different element types seems to be negligible when the force–displacement curves are determined. Hence, especially considering the simulation time, one might perform calculations using T4 elements. Furthermore, when the strain fields obtained from the T4 elements are compared with the DIC measurements in Fig. 11, a very good agreement between the results of experiment and simulation can be observed qualitatively. The statistical results of Pearson's correlation test and the intraclass correlation coefficient also showed a significant correlation between the simulations with the experimental results²¹⁾. This leads to the conclusion that, for clinical practice, the

usage of T4 elements is to be advised.

For the simulation of the prosthesis at chewing velocity, as shown in Fig. 10, the material seems to behave a bit too stiff. However, these observations are most likely due to the simplifications made during the simulation model setup. Particularly because the elasticity of the specimen holder and the metal sphere are not simulated and SPCs are used, a higher overall stiffness for the structure is to be expected. Furthermore, movements and rotations of the antagonist cannot be accounted for, owing to the constrained degrees of freedom of the antagonist, leading to further deviations between the results of simulation and experiment.

The FEA has already been used in numerous areas of dentistry. According to Geng *et al.*²²⁾, Weinstein *et al.*²³⁾ introduced FEA into implantology in 1976. Subsequently, linear FEA was also successfully applied in the fields of prosthetic^{24–26)}, restorative^{27–29)}, endodontic^{30,31)}, and orthodontic^{32,33)} dentistry.

By comparison, nonlinear FEA is rarely used. In the study by Wimmer *et al.*³⁴⁾, for example, certain parameters are reported to behave nonlinearly, but for the purpose of simplification, these parameters are assumed to be linear. The low use of nonlinear FEA can be explained by the extensive material investigations required in comparison to linear FEA²⁾. There are isolated studies in which material properties, such as the nonlinear material behavior of the periodontal ligament, were determined, and a nonlinear simulation was implemented^{11,12,32)}. Nevertheless, numerous studies are available in which the material properties were assumed to be linear, and only contact nonlinearities were considered^{7,35–39)}.

The next challenge and, at the same time, necessity in the use of FEA are validation tests. However, within dental investigations, the calculated results have often been not validated by corresponding *in vitro* tests^{8,28,40,41)}. Furthermore, validation tests cannot always be adequately performed. In the study by Wang *et al.*, experimental *in vitro* tests were conducted with the aid of strain gauges⁴²⁾. However, the strain gauges could not be positioned in the simulation area, and therefore the strain maximum could probably not be recorded. In contrast to the punctual recording of strains using strain gauges, digital image correlation offers the possibility of recording strains on the surfaces of connectors over a large area. For this reason, the simulations in the present study are validated with digital image correlation from a previous study¹⁸⁾. Wang *et al.* stated in their paper that results from FEA calculation models that have not been confirmed by experimental validation tests cannot be considered reliable⁴²⁾.

Following the previous study¹⁸⁾, two test speeds were simulated in the present study. This is much easier to achieve using FEA, in comparison to using classical component testing. However, this requires complete material characterization⁹⁾. The positive possibilities offered by FEA are numerous. Simulations involving different test speeds and load noise are easier and faster to perform than laboratory tests. However, complete

material characterizations are necessary for all FEA simulations because incomplete characterizations can lead to inaccurate simulation results.

The results of the present study reveal that possibly nonlinear behavior of dental biomaterials should be considered in FEA calculations. Extensive material investigations are recommended for material characterization and allow for more accurate simulation results. Furthermore, the present study provides a possible method of simulating the properties of test materials and of minimizing cost-intensive and time-consuming laboratory tests. As already described, it is also possible to simulate different test speeds, especially given that clinical chewing speeds differ significantly from the ISO test velocity⁴³⁾. Nonetheless, the recommended test velocity is very helpful as a comparative variable and for quasi-static movement models (e.g., in bruxism).

A further objective of subsequent studies could be to integrate the nonlinear FEA calculation into an intraoral scanner or CAD/CAM program. In this way, the stability of the digitally designed dental prosthesis could be investigated before the device is manufactured. Weak points could be identified and subsequently reinforced by the user. The final result would be an aesthetic and, at the same time, stability-optimized dental prosthesis. Furthermore, time-consuming and cost-intensive experimental component tests could be avoided in this way.

In follow-up studies, the fracture behavior of dental biomaterials can be further investigated, and an FEA simulation of the fracture can be calculated. This can provide information on whether the location, time, and type of failure can be predicted using an FEA simulation.

In addition, other tests of dental biomaterials such as hybrid composites or CAD/CAM composites could be simulated. In addition, partial or total dentures are also conceivable. For this purpose, however, appropriate reproducible preliminary tests need to be carried out for the respective material. Thereby, the FEA calculation model can then be extended step by step by further parameters, such as the alveolar bone or the periodontal ligament, allowing concrete and accurate clinical statements to be made in the future.

Overall, the use of the semi-analytic model for polymers can be assumed to be reasonable for the given area of use. In this paper, we proposed a methodology for determining all the required material properties, including viscoelastic effects (mainly an increase of elastic modulus with strain rate and stress relaxation) and viscoplastic effects from 3PB experiments. In summary, the results of the present study reveal that it is possible to simulate the behavior of a three-unit bridge restoration. However, for meaningful and valid simulations, expert knowledge in the technical field and extensive preliminary work are both required. Even if the creation of colored illustrations of simulation results is not very complex in comparison, the interpretation of the results is not possible without previous knowledge in the

field of engineering. Otherwise, inaccurate statements and erroneous conclusions may result. This is of great importance for dental biomaterial science, where 3PB tests are standard in material testing. Furthermore, future integration of FEA in intraoral scanners or CAD software is conceivable. These developments would make it possible to provide statements about complex stress conditions and the durability of a prosthesis before it is manufactured.

Moreover, given that different test speeds can lead to different results, a higher test speed comparable to chewing speed should be considered in future material investigations.

CONCLUSIONS

Overall, the following conclusions can be drawn:

1. When a material model for PMMA in dental applications is determined, the influence of stress-state-dependent plastic flow seems to be negligible. The selection of a von Mises yield criterion is deemed sufficiently accurate, leading to a decreased experimental effort.
2. In the case of plastic deformations, the use of the Schmachtenberg's hardening law leads to a good agreement between the experimental and simulation results.
3. Simplifying assumptions of linear elastic material properties for polymeric materials should be avoided in FEA studies, because rate dependencies and plastic flow cannot be modeled. Therefore, precise preliminary investigations for material characterization are necessary.

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