**University of
Zurich**^{UZH}

**Zurich Open Repository and
Archive**

University of Zurich
University Library
Strickhofstrasse 39
CH-8057 Zurich
www.zora.uzh.ch

Year: 2021

Low-Profile Electromagnetic Field Sensors in the Measurement and Modelling of Three-Dimensional Jaw Kinematics and Occlusal Loading

Woodford, Sarah C ; Robinson, Dale L ; Edelmann, Cornelia ; Mehl, Albert ; Röhrle, Oliver ; Vee Sin Lee, Peter ;
Ackland, David C

Abstract: Dynamic occlusal loading during mastication is clinically relevant in the design and functional assessment of dental restorations and removable dentures, and in evaluating temporomandibular joint dysfunction. The aim of this study was to develop a modelling framework to evaluate subject-specific dynamic occlusal loading during chewing and biting over the entire dental arch. Measurements of jaw motion were performed on one healthy male adult using low-profile electromagnetic field sensors attached to the teeth, and occlusal anatomy quantified using an intra-oral scanner. During testing, the subject chewed and maximally compressed a piece of rubber between both second molars, first molars, premolars and their central incisors. The occlusal anatomy, rubber geometry and experimentally measured rubber material properties were combined in a finite element model. The measured mandibular motion was used to kinematically drive model simulations of chewing and biting of the rubber sample. Three-dimensional dynamic bite forces and contact pressures across the occlusal surfaces were then calculated. Both chewing and biting on the first molars produced the highest bite forces across the dental arch, and a large amount of anterior shear force was produced at the incisors and the second molars. During chewing, the initial tooth-rubber contact evolved from the buccal sides of the molars to the lingual sides at full mouth closure. Low-profile electromagnetic field sensors were shown to provide a clinically relevant measure of jaw kinematics with sufficient accuracy to drive finite element models of occlusal loading during chewing and biting. The modelling framework presented provides a basis for calculation of physiological, dynamic occlusal loading across the dental arch.

DOI: <https://doi.org/10.1007/s10439-020-02688-6>

Posted at the Zurich Open Repository and Archive, University of Zurich

ZORA URL: <https://doi.org/10.5167/uzh-195992>

Journal Article

Accepted Version

Originally published at:

Woodford, Sarah C; Robinson, Dale L; Edelmann, Cornelia; Mehl, Albert; Röhrle, Oliver; Vee Sin Lee, Peter; Ackland, David C (2021). Low-Profile Electromagnetic Field Sensors in the Measurement and Modelling of Three-Dimensional Jaw Kinematics and Occlusal Loading. *Annals of Biomedical Engineering*, 49(6):1561-1571.

DOI: <https://doi.org/10.1007/s10439-020-02688-6>

**LOW-PROFILE MAGNETIC FIELD SENSORS IN THE
MEASUREMENT AND MODELLING OF THREE-DIMENSIONAL
JAW KINEMATICS AND OCCLUSAL LOADING**

Sarah C. Woodford¹, Dale L. Robinson¹, Cornelia Edelmann², Albert Mehl², Oliver Röhrle³,
Peter Lee¹, and David C. Ackland¹

¹Department of Biomedical Engineering, University of Melbourne, Parkville, Australia

²Centre of Dental Medicine, University of Zurich, Zurich, Switzerland

³Institute for Modelling and Simulation of Biomechanical Systems, University of Stuttgart,
Germany

Submitted as an original article to *Annals of Biomedical Engineering*

Word count (Including references): 5,783

Address for correspondence:

David C. Ackland

Department of Biomedical Engineering, University of Melbourne

Parkville, Victoria 3010, AUSTRALIA

Phone: +613 8344 0405

Fax: +613 9347 8784

Email: dackland@unimelb.edu.au

ABSTRACT

Dynamic occlusal loading during mastication is clinically relevant in the design and functional assessment of dental restorations and removable dentures, and in evaluating temporomandibular joint dysfunction. The aim of this study was to develop a modelling framework to evaluate subject-specific dynamic occlusal loading during chewing and biting over the entire dental arch. Measurements of jaw motion were performed on [one](#) healthy male adult using low-profile magnetic field sensors attached to the teeth, and occlusal anatomy quantified using an intra-oral scanner. During testing, the subject chewed and maximally compressed a piece of rubber between both second molars, first molars, premolars and their central incisors. [The occlusal anatomy, rubber geometry and experimentally determined rubber material properties were combined to construct a finite element model, where the recorded mandibular motion was used to drive the virtual mandibular teeth into the rubber sample to simulate the experimentally observed chewing and biting.](#) These simulations were used to predict three-dimensional dynamic bite forces and the dynamic contact pressure across the occlusal surfaces. Both chewing and biting on the first molars produced the highest bite forces across the dental arch, and a large amount of anterior shear force was produced at the incisors and the second molars. During chewing, the initial tooth-rubber contact evolved from the buccal sides of the molars to the lingual sides at full mouth closure. [Low-profile magnetic field sensors provide a clinically relevant measure of jaw kinematics with sufficient accuracy to drive finite element models of occlusal loading during chewing and biting.](#) The presented modelling framework provides a basis for calculation of physiological, dynamic occlusal loading across the dental arch.

Key terms: Temporomandibular joint, mandible, motion analysis, biomechanical model,
dental, tooth, occlusion.

INTRODUCTION

Mastication is an essential function required for food breakdown and is achieved by the action of the masticatory muscles in generating dynamic bite force and occlusal loading. The magnitude and direction of bite force, which gives rise to occlusal loading, varies along the dental arch^{3,11} and is influenced by sex⁴⁸, age⁴⁷, facial morphology³¹ and ethnicity⁴³. It is considered an important determinant of jaw function^{3,46}, required for maintaining diet and quality of life²⁷. Measurements of both bite force and occlusal loading have played a role in diagnosing disorders of the temporomandibular joint (TMJ)^{5,23}, assessing mandibular fracture treatment²⁴, and the effects of malocclusions^{3,44}, and are also critical for the design and functional assessment for dental restorations, dental implants and total TMJ prosthetics^{15,33,50}.

Maximum voluntary bite force has been experimentally measured using strain gauges^{11,33}, force transducers⁴⁵, pressure-sensitive film²⁵ and piezoelectric transducers¹³. While these devices may have high force measurement accuracy, they typically provide a static uni-directional force quantity as a subject compresses the device between their teeth, and are inadequate for capturing realistic dynamic occlusal loads such as those occurring during biting, chewing or grinding in situ. Three-component force transducers have also been employed to measure bite force and occlusal shear loading; however, these devices record biting under non-physiological conditions, such as tooth compression onto a metal plate, and may underestimate true bite force magnitude due to the subject's fear of tooth damage during testing¹⁰. Since dynamic force magnitude and direction at the occlusal surface cannot currently be directly measured during normal mastication, the three-dimensional loads incurred by the dental structures and transmitted to the underlying bone remain poorly understood³⁵.

Computational modelling has been used to estimate TMJ loading⁴⁰, prosthetic TMJ stresses¹, dental implant loading³⁴, and the mechanical and thermal loads involved in dental restorations²⁸. Recent modelling studies have estimated dynamic occlusal loading during a maximum force bite (clench) and chewing^{35,36} by employing jaw kinematics data measured experimentally together with detailed models of the occlusal anatomy derived from medical imaging. While this approach produces dynamic compressive and shear force distributions at the occlusal surfaces using finite element model simulations, it requires development of customised dental splints for optical motion analysis as well as micro-computed tomography (μ CT) of dental impressions to digitise occlusal anatomy, which may not be practical in the clinical setting. Furthermore, this approach has only been applied to the first molar to date, and at present, the loading incurred by the second molar and premolars during biting and chewing, and the incisors during dynamic biting, has not been quantified to date.

The aim of this proof-of-concept study was twofold. Firstly, to employ a new low-profile magnetic-field-sensor that can be quickly affixed to dental enamel, together with intra-oral scanning, to produce accurate measurements of occlusal kinematics during biting and chewing; and secondly, to use these measurements to estimate subject-specific occlusal loading over the entire dental arch using computational modelling. The technique reported in the present study provides a strong foundation for practical dynamic kinematics analysis and bite force estimation in both the clinic and research settings.

MATERIALS AND METHODS

Subject recruitment

One male participant (age: 39 yrs, weight: 82 kg) with healthy dentition and no history of mandibular disorders or dental pain was recruited and provided informed consent to participate in this study.

Jaw motion experiments

Motion analysis experiments were performed using the Dental Motion Decoder (DMD) system (Ignident, Ludwigshafen, Germany), which samples at 66 Hz the position and orientation of two low-profile magnetic field sensors of dimensions $1 \times 4 \times 2$ mm ($4 \times 9 \times 3$ mm including coating) within a pre-defined magnetic field volume of approximately 220 cm^3 in size. The measurement error or the sensor position tracking in the capture volume was 0.01 ± 0.20 mm, which was calculated as the error in the distance computed between two adjacent sensors set at a fixed distance of 50.0 mm as they were randomly moved through the entire capture volume (see Supplementary Material). Two low-profile magnetic sensors were fixed to the participant's teeth on the buccal surface using dental adhesive (Optibond FL, Kerr Dental, USA) and cured using ultraviolet light (Demi Plus, Kerr Dental, USA). One sensor was attached to the left maxillary premolar, and another on the right mandibular premolar approximately 3 mm away from the occlusal surface to limit interference during chewing. The participant wore a disposable lip retractor to aid in passage of wires from the mouth, and was seated upright immediately below the magnetic source (Figures 1(a) and (b)). The surfaces of all teeth on the

mandible and maxilla and the magnetic sensors were digitised using an intra-oral scanner (Trios 3 Pod v. 1.18.2.6, 3Shape, Denmark).

The participant then held a soft polyurethane rubber sample of dimensions 20 mm × 20 mm × 5 mm between the second molars on the left (dominant) side, and the rubber sample and second molars were then scanned with the intra-oral scanner to register the position of the rubber relative to the teeth (Figure 1(c)). Ten continuous chewing cycles at the subjects preferred bite force were undertaken. The participant then performed maximal compression of the rubber sample between the second molars for three seconds. The chewing and maximum-force bite protocols were repeated at the first molars and premolars of both left and right sides, as well as the central incisors. During each task, jaw motion was simultaneously recorded from the two magnetic sensors using the DMD system. To register the positions of the sensors to the occlusal anatomy, a triangular surface mesh of the teeth and sensors was generated from the intra-oral scan with an average maximum edge length of 0.15 mm, and the local coordinate system of the mandibular and maxillary sensors expressed in an arbitrary coordinate system associated with the mandibular and maxillary occlusal surface geometries, respectively. Mean inter-occlusal distances were reported for all movements and mean mandibular protrusion and laterotrusion motion projected onto the occlusal plane and reported for the chewing cycles.

Material testing

The material properties of the rubber sample were derived experimentally using a materials test system. The rubber sample was assumed to be a visco-hyperelastic material with a near-incompressible material response, and was represented using a three-element Weichert

model: one non-linear spring element to describe the hyperelastic material behaviour and two parallel Maxwell elements to represent the material's viscoelasticity. The hyperelastic material response was measured using an Instron 5848 Microtester with a 2 kN load cell (Instron, Norwood, MA, USA). Five rubber samples of dimensions 10 mm × 10 mm × 5 mm were compressed at a rate of 0.5 mm per second to a maximum compressive load of 1.5 kN. This load cycle was repeated 3 times and the mean stress-strain curve was used to mathematically derive the hyperelastic material response ([see Supplementary Material](#)). This material behaviour was modelled using the phenomenological Ogden model, which was chosen due to its stability at high compressive strains. For isotropic materials, the strain energy density function is defined as:

$$W(\lambda_1, \lambda_2, \lambda_3) = \sum_{i=1}^N \frac{\mu_i}{\alpha_i} (\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3) \quad (1)$$

where $N = 1$, λ_i are the principle stretch ratios, and μ_i and α_i are the phenomenological material constants ($3.75 \times 10^5 \text{ Pa}$ and $8.35 \times 10^4 \text{ Pa}$, respectively) obtained using least-squares optimisation on the experimentally measured stress-strain curves ($R^2 = 0.9996$). The viscoelastic stress-relaxation response of the five rubber samples was subsequently measured on the Instron 5848 Microtester by performing a compression at a rate of 1 mm per second to a compressive load of 100 N and holding this position for 90 seconds. This stress-relaxation protocol was then repeated for a compressive load of 200 N. The mean relaxation modulus curve was derived from the relaxation modulus of all five specimens, for both the 100 N and 200 N compressions to mathematically model the viscoelastic material response ([see](#)

[Supplementary Material](#)). This viscoelastic model was represented as a second-order Prony series, with a relaxation modulus given by:

$$g_R(t) = 1 - \sum_{i=1}^N g_i (1 - \exp\left(\frac{-t}{\tau_i}\right)) \quad (2)$$

where $N = 2$, t is time and g_i and τ_i are the phenomenological material constants (first order: 2.88×10^{-2} and 0.15 s, second order: 1.71×10^{-2} and 1.78 s respectively), which were obtained by using least-squares optimisation with the experimentally measured stress-relaxation curves ($R^2 = 0.9997$).

Computational modelling

A 3D surface model of the subject's dental geometry was reconstructed from the intra-oral scans using 3-matic 13.0 (Materialise, Leuven, Belgium). The model, which included all maxillary and mandibular teeth, was meshed with triangular shell elements of edge lengths ranging from 0.26 mm to 0.35 mm, and imported into a commercially available finite element modelling package (Abaqus 2017, Dassault Systems, Paris, France).

The three-dimensional geometry of the rubber sample used during the experiment was reconstructed in the finite element model, and intra-oral scans of the participant holding the rubber between their teeth were used to register the position of the virtual rubber sample between the teeth (Figure 1(d)). The virtual rubber sample was meshed using hexahedral elements with an element length of 0.35 mm, and assigned the visco-hyperelastic material properties derived from the mechanical testing. Since tooth enamel has a much greater stiffness compared to that of the rubber, the teeth were assumed rigid bodies.

The mandibular kinematics data obtained from the jaw motion experiments was used to kinematically drive the finite element model simulations of chewing and biting. The measured movement of the mandible corresponded to the movement of the virtual mandibular teeth and was applied as six boundary conditions in the model i.e., three rotations and three translations. The virtual maxillary teeth were assumed fixed, and during simulations, the virtual rubber sample was compressed between the mandibular and maxillary teeth. Tooth-rubber contact was modelled using a general contact formulation with a “hard” normal contact and a “rough” tangential contact law applied to avoid sliding of the teeth over the rubber, which was representative of the experimental test conditions. Mass scaling was applied to select elements of the rubber sample to reduce computational time (see Supplementary Material).

Simulations were performed with parallelization over 6 cores on a desktop PC with an Intel Core i7 3.4 GHz processor and 16 GB memory. Contact pressure was reported over the occlusal surfaces, and bite forces were reported for the mandibular and maxillary teeth with respect to an anatomic co-ordinate system aligned with the occlusal plane. Specifically, the x-, y- and z-directions corresponded to the lateral (left), posterior and superior directions, respectively. Chewing cycles were defined as commencing just after initial contact between the teeth and the rubber (0%) and finishing at the time of maximum bite force (100%).

Model validation

To validate the finite element model, an independent compression experiment was performed on a single tooth against the rubber using the Instron 5848 microtester (Figure 2(a)). One intact, adult second molar was sourced from the Melbourne Dental School and fixed to a

custom-designed potting fixture in dental plaster. This tooth was aligned to a thin cylinder of the rubber material (diameter: 30 mm, height: 5.5 mm), and compressed at a rate of 0.5 mm per second to a displacement of 4 mm, while uniaxial reaction force was measured by the load cell. A finite element model of this experiment was simulated using a cylinder matching the dimension of the rubber and meshed with hexahedral elements of element size 0.35 mm. The tooth-potting fixture assembly was imaged using μ CT (μ CT 50; Scanco Medical, Brüttisellen, Switzerland) with a voxel size of 0.02 mm, and the resultant images were segmented to obtain a surface mesh of the combined fixture and tooth. The geometries were imported into the finite element software and meshed with triangular shell elements of length ranging from 0.23 mm to 0.35 mm. The tooth was modelled as a rigid body and the rubber as a deformable visco-hyperelastic material with material properties and contact conditions as described above. The virtual tooth was compressed against the rubber following the compressive displacement measurements, and the calculated uniaxial reaction force was compared to that measured experimentally.

A mesh convergence analysis was then conducted to evaluate the effect of different virtual rubber sample element sizes on the calculated load cell response. A reduction in element size from 0.35 mm to 0.3 mm was found to decrease the errors introduced both through solving the contact formulation and mesh discretisation by 0.7%; however, this was at the expense of increased computational time.

RESULTS

The model validation simulations demonstrated an RMS error of 1.39 N between the simulated and experimentally measured uniaxial reaction forces on the Instron load cell (Figure 2(b)).

During chewing, the inter-occlusal gap was smallest for the left second molar (1.28 mm) followed by the premolars (left: 1.55 mm, right: 1.51 mm), incisors (1.66 mm), first molars (left: 2.50 mm, right: 2.22 mm) and the right second molar (2.61 mm) (Figure 3). Throughout the mean chewing cycles, the mandible protruded forwards during mouth closure, with the position of maximum mandibular protrusion occurring at 100% mouth closure for both second molars (left: 1.63 mm, right: 2.76 mm), the right first molar (2.13 mm) and the incisors (0.97 mm). Maximum mandibular protrusion during chewing also occurred in the vicinity of 100% mouth closure for the right (2.24 mm) and left (1.75 mm) premolars respectively, while a small amount of mandibular retrusion occurred for the mean chewing cycle on the left first molar (0.3 mm) (Figure 4). Limited mandibular laterotrusion was recorded for all teeth during chewing (Figure 4).

The first molars generated the largest bite forces of all teeth, a finding consistent for both chewing (left: 309 N, right: 311 N) and maximum-force biting (left: 496 N, right: 495 N). The incisors generated the smallest bite forces during chewing (75 N) and maximum-force biting (114 N) (Figure 5). The average bite forces on the first molars, premolars and incisors were 124%, 50% and 33% of that on the second molars (Figure 5, Tables 1 and 2). During chewing, the highest bite forces on the mandibular teeth were directed anteriorly, this was particularly prominent on the incisors, with the maximum force directed 26.7° anteriorly, and the second molars, with the maximum force directed 9.6° and 10.6° anteriorly (Table 1). This

trend was also seen for maximum-force biting, with the maximum force directed 21.4° anteriorly for the incisors and 8.7° and 8.6° anteriorly for the second molars (Table 2). The mandibular bite forces were higher for the maximum-force bite than during chewing across the entire dental arch (Figure 5, Tables 1 and 2).

For the first molar, the tooth-rubber contact evolution during chewing was shown to progress from the buccal sides of both maxillary and mandibular first molars towards the lingual side during jaw closure. At 100% jaw closure during chewing, maximum stress regions occurred over the distobuccal and distolingual cusps of the first molars, while the highest stresses in the rubber occurred between the disto-lingual cusps of the maxillary and mandibular. During chewing with the second molars, the tooth-rubber contact was initiated on the buccal side of the mandibular tooth, and the buccal side of the mesial maxillary cusp and the lingual side of the posterior maxillary cusp, this progressed towards the lingual side of both second molars during jaw closure (Figure 6). By 100% jaw closure, the highest stress in the rubber was between the maxillary mesiobuccal cusp and its opposing mandibular ridge of the second molars.

DISCUSSION

This study presented a subject-specific computational modelling framework for evaluating jaw kinematics, bite forces and the occlusal loading distribution during chewing and maximum-force biting using a new magnetic-field-based kinematics measurement system. In conventional approaches, jaw kinematics are measured using mechanical devices such as

axiography³⁹, point tracking methods such as optoelectronic tracking⁶ or radiographic techniques such as x-ray fluoroscopy⁸. Mechanical devices and point tracking techniques often involve attachment of bulky equipment such as face-bows to the mandibular teeth, which may load the jaw inferiorly and interfere with natural jaw motion and occlusion. Additionally, techniques such as optoelectronic tracking typically use skin mounted markers, which are prone to significant skin motion artefact⁷. Radiographic methods such as 4D Computed Tomography, which provide direct kinematic measurement without skin motion artefact, expose the subject to ionising radiation, and have therefore had limited application. [For a full review of mandibular motion tracking techniques see Woodford et al. \(2020\)⁵¹](#). In contrast, the DMD system employed in the present study provides accurate non-invasive jaw kinematics measurement that can be rapidly performed chairside with little setup time. The low-profile electromagnetic trackers that are adhered to the teeth do not interfere with natural motion of the jaw during chewing, [they provide dynamic measurement of kinematics without skin motion artefact, and bypass the requirement to fabricate subject-specific splints](#). These features lend the system well to jaw motion measurement in both the research laboratory and clinic.

[Traditional bite force measurements are limited to quantifying static, unidirectional, maximal occlusal loads obtained through non physiological conditions such as clenching on a hard metal surface^{22,35}](#). Whilst previous efforts have been made to measure the three-dimensional components of occlusal loads, this technology relies on implanting sensors within restored teeth and therefore cannot be applied to participants with healthy dentition^{20,29}. Similarly, dynamic occlusal loading has been measured using a novel application of strain gauges, however these recordings are also limited to quantification of unidirectional loads⁴².

The presented measurement and modelling methodology provides a novel alternative to existing measurement techniques, it does not require the participant to bite down on a hard surface, which has been shown to prevent maximum bite force capacity²². To the authors knowledge this is the first technique capable of measuring dynamic three-dimensional loading and thus allows the analysis of sub-maximal loading of the teeth, like that observed in functionally relevant activities such as mastication, and permits quantification of shear loading at the tooth surface, which negatively affects the stress distribution within dental structures³⁵. This modelling methodology overcomes practical limitations associated with bite-force measurements, and shows great promise for research in the fields of biomechanics and dental restorations.

The results of this study are in reasonable agreement with data reported sporadically in the literature. For example, the superior, protrusive and lateral mandibular movements during chewing with the first molar (5.9 mm, 2.2 mm and 0.54 mm respectively) reported in the present study were similar to those of Röhrle et al. (2018) who also studied chewing of rubber samples with similar material properties and dimensions (approximately 5.0 mm, 2.0 mm and 0.50 mm respectively)³⁶. This study found maximum-force bite loads of approximately 400 N, 495 N, 195 N and 115 N for the second molars, first molars, premolars and central incisors respectively. These results are consistent with bite force data previously measured for the first molars (546 N)³², premolars (254 N) and central incisors (146 N)¹¹, but are lower than some average bite forces reported elsewhere for healthy adult males^{2,10,12,18,46–48} (Table 3). This study found chewing loads of approximately 249 N, 310 N, 127 N and 75 N for the second molars, first molars, premolars and central incisors respectively. These values are lower than those reported

for cyclic chewing of rubber on the first molar and premolars (617.8 N)³⁶, which may be due to differences in the models used to represent the rubber material properties. However, we surmise that the calculated bite forces fall within the expected range given the large variability in bite force measurements between subjects and across different measurement modalities⁴⁵.

The highest occlusal forces were consistently located at the first molars, both for chewing and maximum-force biting. This finding contradicts the mechanical lever theory¹⁴, a popular belief that the jaw functions as a third-class lever. In this model, the muscle produces a force located between the condyle, which acts as a hinge, and a tooth which produces a bite force. Based on this theory, for a constant maximum muscle force, the magnitude of the maximum occlusal load applied by a tooth depends only on the distance between the tooth and the muscle attachment point on the mandible. Therefore, the posterior teeth closer to the muscle attachment point are capable of higher occlusal loads than anterior teeth located further from the muscle attachment point. The mechanical lever theory predicts that the bite forces on the first molar, premolars and central incisors will be 90%, 83% and 69% of that on the second molar respectively¹⁴, however, the results of this study found the bite forces on the first molar, premolars and central incisors to be 124%, 50% and 33% of that on the second molars, respectively, while several other empirical studies reported similar inconsistencies with the mechanical lever theory^{11,16,17}. Our finding supports the alternative principle stated by Hattori et. al. that the critical limit of maximum jaw loading may be the load-bearing capacity of the periodontal tissues¹⁷ – since the first molar generally has the largest root surface area of all teeth¹⁹, less load is transmitted to the periodontal tissue and larger bite forces may be permissible without pain. The findings of this study also suggest that the periodontally borne

load, expressed as tooth-specific bite force divided by the total tooth root surface area, is constant across the dental arch¹⁷.

This study demonstrated that a large amount of anterior shear loading is present during chewing and maximal-force biting, particularly on the incisors and second molars, a finding consistent with other studies of three-dimensional occlusal loading^{16,36,38}. During chewing and maximum-force biting the subject's mandible moved anteriorly with little lateral motion, resulting in more anterior shear loading than lateral shear loading. Shear loading at the tooth surface greatly increases the magnitude of the maximum stress within the tooth³⁷ and introduces a bending moment at the bone near the base of the tooth due to the shear loading at the occlusal surface producing a tooth cantilever effect³⁶. For implant supported prostheses, this could increase the transfer of occlusal loads to the underlying cortical bone and may lead to negative outcomes such as early implant loosening or failure of the implant-bone interface⁴. Additionally, the higher stresses at either an implanted or restored tooth's surface may lead to implant failure due to fractures or fatigue failure⁹. Furthermore, the production of non-axial occlusal loads requires higher activation forces from the masticatory muscles and results in higher TMJ loads in comparison to axial occlusal loads^{38,41}. The results of this study indicate that musculoskeletal model predictions of TMJ loading that use bite force measures with no shear loading considerations may ultimately under-estimate joint force predictions.

The present study showed that the evolution of occlusal contact during chewing is consistent with studies of the first molar and premolars that found that the bolus is compressed between opposing cusps of the maxillary and mandibular teeth³⁶. This interaction between opposing cusps may be necessary to achieve the force and stress concentrations necessary to

grind and breakdown food during mastication, which is supported by evidence that subjects with greater occlusal contact area are better equipped to mechanically breakdown foods^{30,49}.

This study showed that the mean inter-occlusal gap during chewing on the left second molar (1.28 mm) was much smaller than that of its neighbouring first molar (2.50 mm) or the right second molar (2.61 mm) (Figure 3). We note that variability between the repeated occlusal movements may be present in these results, potentially arising from subject familiarisation with the experimental protocol. This variability has been found for both within- and between-cycle recordings, and is largest for the vertical displacement measures^{21,26}. In this experiment, the left second molar was the first loading trial to be conducted – the unexpectedly low inter-occlusal gap may be due to subject adaption to the experimental protocol, which should be considered in future studies.

Some limitations of this study ought to be noted. First, errors in kinematics measurement translate directly to variability in occlusal loading, but the extent of this is not well understood. We ran four finite element simulations of habitual mastication for the right first molar with the mandibular jaw closure offset by the mean measurement error of ± 0.01 mm, as well as mean plus standard deviation of ± 0.21 mm, which produced force calculation errors of ± 2 N and ± 42 N, respectively (see Supplementary Material). This demonstrates that calculation of bite force using finite element model simulations in the present study depend on highly accurate measurements of jaw kinematics, as well as high quality registration of magnetic sensors to occlusal anatomy. Second, errors in the calculated occlusal load may also be introduced through the material model, contact formulation, mesh discretisation and the solver used in the finite element analysis, however the mesh convergence and validation

studies were performed to keep these errors to a minimum whilst retaining a viable computational expense. Third, this study was limited to a sample size of one, and future studies ought to employ this technique on a larger participant cohort to enable generalisation of results and a thorough statistical analysis.

In conclusion, this study reports a new high accuracy method of measuring jaw kinematics using low-profile magnetic field sensors that are suitable for both the clinic and research settings. By combining these measurements with occlusal anatomy derived from intra-oral scans and finite element model simulations, we derived calculations of dynamic, three-dimensional occlusal loading during biting and chewing. This measurement and modelling framework, which we validated with independent testing, produces physiologically relevant occlusal load estimates which are otherwise impossible to achieve using conventional bite force transducers. This study showed that unilateral chewing and biting using the first molars produced more bite force than that of the second molar, pre-molar and incisors, supporting the theory that the critical limit of maximum jaw loading may be the load-bearing capacity of the periodontal tissues. Future studies ought to employ this technique on a larger participant cohort to enable generalisation of results. The motion measurement system and modelling framework presented may be useful for design and evaluation of patient-specific prosthodontic devices and dental restorations, diagnosis of TMJ disorders, and in surgical planning of dental, oral and maxillofacial surgical procedures.

BIBLIOGRAPHY

1. Ackland, D. C., A. Moskaljuk, C. Hart, P. Vee Sin Lee, and G. Dimitroulis. Prosthesis Loading after Temporomandibular Joint Replacement Surgery: A Musculoskeletal Modeling Study. *J. Biomech. Eng.* 137:, 2015.
2. Al-Omiri, M. K., M. G. Sghaireen, M. M. Alhijawi, I. A. Alzoubi, C. D. Lynch, and E. Lynch. Maximum bite force following unilateral implant-supported prosthetic treatment: within-subject comparison to opposite dentate side. *J. Oral Rehabil.* 41:624–629, 2014.
3. Bakke, M. Bite Force and Occlusion. *Semin. Orthod.* 12:120–126, 2006.
4. Barbier, L., J. Vander Sloten, G. Krzesinski, E. Schepers, and G. Van Der Perre. Finite element analysis of non-axial versus axial loading of oral implants in the mandible of the dog. *J. Oral Rehabil.* 25:847–858, 1998.
5. Bonjardim, L. R., M. B. D. Gaviao, L. J. Pereira, and P. M. Castelo. Bite force determination in adolescents with and without temporomandibular dysfunction. *J. Oral Rehabil.* 32:577–583, 2005.
6. Buschang, P. H., H. Hayasaki, and G. S. Throckmorton. Quantification of human chewing-cycle kinematics. *Arch. Oral Biol.* 45:461–474, 2000.
7. Chen, C.-C., Y.-J. Chen, S.-C. Chen, H.-S. Lin, and T.-W. Lu. Evaluation of soft-tissue artifacts when using anatomical and technical markers to measure mandibular motion. *J. Dent. Sci.* 6:95–101, 2011.
8. Chen, C.-C., C.-C. Lin, T.-W. Lu, H. Chiang, and Y.-J. Chen. Feasibility of differential

- quantification of 3D temporomandibular kinematics during various oral activities using a cone-beam computed tomography-based 3D fluoroscopic method. *J. Dent. Sci.* 8:151–159, 2013.
9. Drummond, J. L. Degradation, Fatigue, and Failure of Resin Dental Composite Materials. *J. Dent. Res.* 87:710–719, 2008.
 10. van Eijden, T. M. G. J. Three-Dimensional Analysis of Human Bite-Force Magnitude and Moment. *Arch. Oral Biol.* 36:535–539, 1991.
 11. Ferrario, V. F., C. Sforza, G. Serrao, C. Dellavia, and G. M. Tartaglia. Single tooth bite forces in healthy young adults. *J. Oral Rehabil.* 31:18–22, 2004.
 12. Ferrario, V. F., C. Sforza, G. Zanotti, and G. M. Tartaglia. Maximal bite forces in healthy young adults as predicted by surface electromyography. *J. Dent.* 32:451–457, 2004.
 13. Flanagan, D., H. Ilies, B. O’Brien, A. McManus, and B. Larrow. Jaw Bite Force Measurement Device. *J. Oral Implantol.* 38:361–364, 2012.
 14. Gosen, A. J. Mandibular leverage and occlusion. *J. Prosthet. Dent.* 31:369–376, 1974.
 15. Hasan, I., C. Madarlis, L. Keilig, C. Dirk, A. Weber, C. Bourauel, and F. Heinemann. Changes in biting forces with implant-supported overdenture in the lower jaw: A comparison between conventional and mini implants in a pilot study. *Ann. Anat.* 208:116–122, 2016.
 16. Hattori, Y., C. Satoh, T. Kunieda, R. Endoh, H. Hisamatsu, and M. Watanabe. Bite forces and their resultants during forceful intercuspal clenching in humans. *J. Biomech.*

42:1533–1538, 2009.

17. Hattori, Y., C. Satoh, S. Seki, Y. Watanabe, Y. Ogino, and M. Watanabe. Occlusal and TMJ Loads in Subjects with Experimentally Shortened Dental Arches. *J. Dent. Res.* 82:532–536, 2003.
18. Hellsing, G. On the regulation of interincisor bite force in man. *J. Oral Rehabil.* 7:403–411, 1980.
19. Hujoel, P. P. A meta-analysis of normal ranges for root surface areas of the permanent dentition. *J. Clin. Periodontol.* 21:225–229, 1994.
20. Kawaguchi, T., T. Kawata, T. Kuriyagawa, and K. Sasaki. In vivo 3-dimensional measurement of the force exerted on a tooth during clenching. *J. Biomech.* 40:244–251, 2007.
21. Keeling, S. D., C. H. Gibbs, S. M. Lupkiewicz, G. J. King, and A. P. Jacobson. Analysis of repeated-measure multicycle unilateral mastication in children. *Am. J. Orthod. Dentofac. Orthop.* 99:402–408, 1991.
22. Koc, D., A. Dogan, and B. Bek. Bite Force and Influential Factors on Bite Force Measurements: A Literature Review. *Eur. J. Dent.* 4:223–232, 2010.
23. Kogawa, E. M., P. S. Calderon, J. R. P. Lauris, C. R. P. Araujo, and P. C. R. Conti. Evaluation of maximal bite force in temporomandibular disorders patients. *J. Oral Rehabil.* 33:559–565, 2006.
24. Kshirsagar, R., N. Jaggi, and R. Halli. Bite Force Measurement in Mandibular

- Parasymphyseal Fractures: A Preliminary Clinical Study. *Craniomaxillofac. Trauma Reconstr.* 4:241–244, 2011.
25. Kumagai, H., T. Suzuki, T. Hamada, P. Sondang, M. Fujitani, and H. Nikawa. Occlusal force distribution on the dental arch during various levels of clenching. *J. Oral Rehabil.* 26:932–935, 1999.
 26. Lassauzay, C., M.-A. Peyron, E. Albuissou, E. Dransfield, and A. Woda. Variability of the masticatory process during chewing of elastic model foods. *Eur. J. Oral Sci.* 108:484–492, 2000.
 27. Lepley, C. R., G. S. Throckmorton, R. F. Ceen, and P. H. Buschang. Relative contributions of occlusion, maximum bite force, and chewing cycle kinematics to masticatory performance. *Am. J. Orthod. Dentofac. Orthop.* 139:606–613, 2011.
 28. Mackerle, J. Finite Element Modelling and Simulations in Dentistry : A Bibliography 1990 – 2003. *Comput. Methods Biomech. Biomed. Engin.* 7:277–303, 2004.
 29. Mericske-stern, R. Three-Dimensional Force Measurements With Mandibular Overdentures Connected to Implants by Ball-Shaped Retentive Anchors . A Clinical Study. *Int. J. Oral Maxillofac. Implant.* 13:36–43, 1998.
 30. Owens, S., P. H. Buschang, G. S. Throckmorton, L. Palmer, and J. English. Masticatory performance and areas of occlusal contact and near contact in subjects with normal occlusion and malocclusion. *Am. J. Orthod. Dentofac. Orthop.* 121:602–609, 2002.
 31. Proffit, W. R., H. W. Fields, and W. L. Nixon. Occlusal Forces in Normal- and Long-face

- Adults. *J. Dent. Res.* 62:566–570, 1983.
32. Raadsheer, M. C., T. M. G. J. van Eijden, F. C. van Ginkel, and B. Prahl-Andersen. Contribution of Jaw Muscle Size and Craniofacial Morphology to Human Bite Force Magnitude. *J. Dent. Res.* 78:31–42, 1999.
 33. Rismanchian, M., F. Bajoghli, Z. Mostajeran, A. Fazel, and P. Eshkevari. Effect of implants on maximum bite force in edentulous patients. *J. Oral Implantol.* 35:196–200, 2009.
 34. Robinson, D., L. Aguilar, A. Gatti, J. Abduo, P. V. . Lee, and D. Ackland. Load response of the natural tooth and dental implant: A comparative biomechanics study. *J. Adv. Prosthodont.* 11:169–178, 2019.
 35. Röhrle, O., H. Saini, and D. C. Ackland. Occlusal loading during biting from an experimental and simulation point of view. *Dent. Mater.* 34:58–68, 2018.
 36. Röhrle, O., H. Saini, P. V. S. Lee, and D. C. Ackland. A novel computational method to determine subject-specific bite force and occlusal loading during mastication. *Comput. Methods Biomech. Biomed. Engin.* 21:453–460, 2018.
 37. Romeed, S. A., R. Malik, and S. M. Dunne. Stress Analysis of Occlusal Forces in Canine Teeth and Their Role in the Development of Non-Carious Cervical Lesions: Abfraction. *Int. J. Dentistry* 153–159, 2012.doi:10.1155/2012/234845
 38. Rues, S., H. J. Schindler, J. C. Türp, K. Schweizerhof, and J. Lenz. Motor behavior of the jaw muscles during different clenching levels. *Eur. J. Oral Sci.* 116:223–228, 2008.
 39. Sadat-Khonsari, R., C. Fenske, B. Kahl-Nieke, I. Kirsch, and H. D. Jüde. The helical axis of

- the mandible during the opening and closing movement of the mouth. *J. Orofac. Orthop.* 64:178–185, 2003.
40. Sagl, B., M. Schmid-Schwap, E. Piehslinger, M. Kundi, and I. Stavness. A Dynamic Jaw Model With a Finite-Element Temporomandibular Joint. *Front. Physiol.* 10:1–12, 2019.
 41. Schindler, H. J., S. Rues, J. C. Türp, K. Schweizerhof, and J. Lenz. Jaw clenching: muscle and joint forces, optimization strategies. *J. Dent. Res.* 86:843–847, 2007.
 42. Shimada, A., Y. Yamabe, T. Torisu, L. Baad-Hansen, M. H, and P. Svensson. Measurement of dynamic bite force during mastication. *J. Oral Rehabil.* 39:349–356, 2012.
 43. Shinogaya, T., M. Bakke, C. E. Thomsen, A. Vilmann, A. Sodeyama, and M. Matsumoto. Effects of ethnicity, gender and age on clenching force and load distribution. *Clin. Oral Investig.* 5:63–68, 2001.
 44. Sonnesen, L., M. Bakke, and B. Solow. Bite force in pre-orthodontic children with unilateral crossbite. *Eur. J. Orthod.* 23:741–749, 2001.
 45. Tortopidis, D., M. F. Lyons, R. H. Baxendale, and W. H. Gilmour. The variability of bite force measurement between sessions, in different positions within the dental arch. *J. Oral Rehabil.* 25:681–686, 1998.
 46. Umesh, S., S. Padma, S. Asokan, and T. Srinivas. Fiber Bragg Grating based bite force measurement. *J. Biomech.* 49:2877–2881, 2016.
 47. Varga, S., S. Spalj, M. L. Varga, S. A. Milosevic, S. Mestrovic, and M. Slaj. Maximum voluntary molar bite force in subjects with normal occlusion. *Eur. J. Orthod.* 33:427–433,

2011.

48. Waltimo, A., and M. Kononen. A novel bite force recorder and maximal isometric bite force values for healthy young adults. *Scandinavian J. Dent. Res.* 101:171–175, 1993.
49. Wilding, R. J. C. The Association Between Chewing Efficiency and Occlusal Contact Area in Man. *Arch. Oral Biol.* 38:589–596, 1993.
50. Winter, A., A. Schurig, E. Rasche, F. Rosner, L. Kanus, and M. Schmitter. The flexural strength of CAD/CAM polymer crowns and the effect of artificial ageing on the fracture resistance of CAD/CAM polymer and ceramic single crowns. *J. Mater. Sci. Mater. Med.* 31:, 2020.
51. Woodford, S. C., D. L. Robinson, A. Mehl, P. V. S. Lee, and D. C. Ackland. Measurement of normal and pathological mandibular and temporomandibular joint kinematics: A systematic review. *J. Biomech.* 111:109994, 2020.

FIGURE CAPTIONS

- Figure 1: a) Electromagnetic sensors glued to the mandibular and maxillary premolar, with lip and cheek retractors to aid passage of sensor wires and avoid contact with sensors. b) DMD experimental setup showing participant with magnetic field sensors attached to the teeth and the magnetic source located approximately 8 cm above the subject's head. c) 3D model of the participant's mandibular and maxillary teeth with rubber held between the premolars obtained through the intra-oral scan, the red box indicates the region that was input into the finite element model. d) Finite element model of the premolars (grey) and rubber (blue) constructed from the intra-oral scan data. Teeth that are not in contact with the rubber have been removed for computational efficiency.
- Figure 2: Validation experimental setup showing human tooth cemented into a custom-made fixture, the blue rubber sample lies between the tooth and the load cell (a) and experimental results compared against computational model output corresponding to this loading scenario (b). The RMS error between experiment and model was 1.39 N.
- Figure 3: Vertical translation of the mandible measured during ten chewing cycles. Black curves represent mean cycle data, while the shaded region indicates one standard deviation about the mean.

- Figure 4: Mandibular protrusive and laterotrusive movements during mean chewing of the rubber sample, projected onto the occlusal plane. Data is given for the incisors, premolars, first molars and second molars. White, yellow and black markers refer to the points at 0%, 50% and 100% of the chewing cycle respectively. x, y and z represent the lateral (left), posterior and superior directions respectively. All trajectory images adhere to scale in lower left corner.
- Figure 5: Tooth-specific mandibular bite forces for mean chewing and maximum-force bite tests. Data is given for the incisors, premolars, first molars and second molars. All forces are given in Newtons; x, y and z represent the lateral (left), posterior and superior directions, respectively.
- Figure 6: Evolution of the contact pressure distribution along the maxillary (top row) and mandibular (bottom row) dental arches at 25%, 50%, 75% and 100% of the mean chewing cycle. Contact pressure values are given in MPa.

TABLE CAPTIONS

- Table 1: Magnitude and direction of the mean chewing force for each mandibular tooth. Force values are provided at 25%, 50%, 75% and 100% of the chewing cycle. Directions are given in degrees with respect to the sagittal plane (X) and the frontal plane (Y).
- Table 2: Magnitude and direction of the peak maximum-force bite value for each mandibular tooth. Directions are given in degrees with respect to the sagittal plane (X) and the frontal plane (Y).
- Table 3: Published tooth-specific maximum bite forces for adult males with healthy dentition. Data includes gender, age and sample size of subjects, and the reported bite forces for the second molars, first molars, premolars and incisors, where available.

Figure 1

Figure 2

(a)

(b)

Figure 3

Figure 4

Figure 5

Figure 6

Table 1

Chewing								
Side	Tooth	Cycle progression	Bite force (N)				Bite force direction (degrees)	
			X	Y	Z	Magnitude	X	Y
Left	Second molar	25%	-0.96	-11.00	43.82	45.19	-1.26	-14.10
		50%	-3.56	-20.32	83.46	85.97	-2.44	-13.68
		75%	-8.13	-31.40	152.81	156.22	-3.05	-11.61
		100%	-31.40	-49.23	291.84	297.62	-6.14	-9.57
	First molar	25%	-2.77	-5.90	41.39	41.90	-3.82	-8.12
		50%	-6.37	-12.18	86.55	87.63	-4.21	-8.01
		75%	-13.98	-20.70	170.30	172.13	-4.69	-6.93
		100%	-28.16	-31.42	305.82	308.72	-5.26	-5.87
	Premolar	25%	-3.69	-1.72	20.57	20.97	-10.17	-4.77
		50%	-7.12	-3.11	41.67	42.39	-9.69	-4.27
		75%	-13.59	-4.75	76.98	78.31	-10.01	-3.53
		100%	-23.86	-7.29	131.63	133.97	-10.28	-3.17
Central	Incisor	25%	0.10	-4.73	11.47	12.41	0.50	-22.40
		50%	-0.64	-11.59	26.60	29.02	-1.37	-23.54
		75%	-1.38	-20.91	44.69	49.36	-1.76	-25.07
		100%	-2.18	-33.77	67.40	75.41	-1.85	-26.61
Right	Premolar	25%	0.47	-1.85	24.85	24.92	1.08	-4.26
		50%	0.21	-3.58	45.63	45.77	0.26	-4.49
		75%	1.19	-5.66	78.09	78.30	0.88	-4.14
		100%	3.37	-8.23	120.30	120.63	1.60	-3.91
	First molar	25%	-2.04	-7.56	50.00	50.61	-2.25	-8.63
		50%	-3.80	-14.21	101.15	102.21	-2.12	-8.21
		75%	-3.31	-23.17	176.65	178.19	-1.14	-7.26
		100%	-0.16	-34.08	308.87	310.74	-0.03	-6.30
	Second molar	25%	1.80	-8.03	40.64	41.47	2.53	-11.17
		50%	4.17	-15.56	75.15	76.86	3.17	-11.70
		75%	8.87	-25.59	124.04	126.96	4.09	-11.66
		100%	15.00	-36.52	196.00	199.93	4.38	-10.55

Table 2

Maximum bite							
Side	Tooth	Bite force (N)				Bite force direction (degrees)	
		X	Y	Z	Magnitude	X	Y
Left	Second molar	-53.82	-62.45	406.72	415.00	-7.56	8.70
	First molar	-37.39	-45.50	492.67	496.18	-4.62	5.19
	Premolar	-37.88	-13.54	198.39	202.43	-10.97	4.09
Central	Incisor	-0.22	-41.63	106.50	114.34	-0.12	21.35
	Premolar	14.09	-3.48	183.58	184.20	4.62	1.16
Right	First molar	15.88	-40.93	493.04	495.00	1.74	4.62
	Second molar	28.20	-56.22	380.20	385.37	4.04	8.62

Table 3

Gender	Age (years)	Sample Size	Bite Force (mean $\pm$ SD) (N)				Authors	Reference
			Second Molar	First Molar	Premolar	Incisors		
M	na	8	-	-	-	161 $\pm$ 29.6	Hellsing, 1980	18
M	31.1 (mean)	na	723 $\pm$ 138	-	583.0 $\pm$ 99.0	-	van Eijden, 1991	10
M	22-35	15	-	847.0 $\pm$ 131.0	-	287.0 $\pm$ 75.0	Waltimo and Kononen, 1993	48
M	18-36	58	-	545.7 $\pm$ 115.1	-	-	Raadsheer et al., 1999	32
M	20-29	8	-	785.0 $\pm$ 203.1	-	-	Ferrario et al., 2004	12
M	19-29	36	294.3 $\pm$ 55.92	306.1 $\pm$ 42	254.1 $\pm$ 72.2	146.17 $\pm$ 44.4	Ferrario et al., 2004	11
M	18	14	-	777.7 $\pm$ 78.7	-	-	Varga et al., 2011	47
M	28-57	20	-	651.5 $\pm$ 59.9	-	-	Al-Omiri et al., 2014	2
M	24-35	20		629.5 $\pm$ 64.5	365.0 $\pm$ 36.0	262.5 $\pm$ 32.0	Umesh et al., 2016	46

SUPPLEMENTARY MATERIAL

The precision of the DMD sensor position tracking was verified by calculating the distance between two sensors set on a custom-designed 3D-printed plate that rigidly cradled the sensors, separating them with a distance of 50.00 mm. The plate with the sensors attached was moved through the entire capture volume of the DMD system over approximately 30 seconds, and the difference between the calculated and actual sensor spacing was computed and used as a measure of the system precision (Figure S1).

Figure S1: DMD precision, evaluated from differences between calculated and actual sensor spacing as two sensors glued to a plate were moved through the magnetic field capture volume. The colour bar indicates the range of position measurement error in millimetres. The actual sensor spacing was 50.00 mm. Displayed are a sample of the complete trajectories used in the calculation.

All rubber samples used in this study were from the same piece of commercial extruded rubber. They exhibited some variations in their material properties both between- and within-samples. To capture the average behaviour of the rubber, 10 different samples were tested (5 for stress-strain tests and 5 for stress-relaxation tests) and the average results used to fit material models to. The variations in the experimental material response and average curves used for model fitting can be seen in figures S2.

Figure S2: Experimental polyurethane rubber material responses used for characterising rubber material properties. (a) Stress-strain response used for calculating hyperelastic material response. (b) Stress-relaxation curves (100N) used for calculating viscoelastic response. (c) Stress-relaxation curves (200N) used for calculating viscoelastic response. Note, that the average of the 100N and 200N stress-relaxation curves were used for fitting the prony series to.

The phenomenological Ogden model was chosen to model the material's hyperelastic response for its good stability at high strain ranges similar to those seen during the biting experiment. A second order prony model was used to model the materials viscoelastic response, this was tuned to model the initial 3 seconds of the stress relaxation response as this was the longest time frame that the rubber was held in a constant position during the biting

experiments. The hyperelastic and viscoelastic material model fit to the experimental data can be seen in figure S3.

Figure S3: Material model fit to experimental data. (a) Hyperelastic material response, fit of first order phenomenological Ogden model to experimental average stress-strain data. (b) Viscoelastic material response, fit of second order Prony series to experimental average stress-relaxation results.

Mass scaling was applied to the simulations to reduce the computation time. To assess whether this addition still lead to a meaningful solution the maximum ratio of kinetic energy to internal energy was calculated as a percentage for all simulations to ensure that inertial forces remained insignificant throughout the simulations. All values remain below the accepted 2% (Table S1).

Table S1: Maximum ratio of kinetic energy to internal energy for each simulation.

	Tooth	Chewing	Maximal-force biting
Left	Second molar	0.11 %	0.03 %
	First molar	0.04 %	0.01 %
	Premolar	0.97 %	0.08 %
Central	Incisor	0.29 %	0.04 %
Right	Premolar	0.10 %	0.03 %
	First molar	0.06 %	0.01 %
	Second molar	0.09 %	0.01 %

The effect of the propagated errors from the kinematic measurement system on the calculated bite force values was analysed by running 4 separate simulations of mean chewing on the right 1st molar with the mandible offset by the mean measurement error of ± 0.01 mm, as well as mean plus standard deviation of ± 0.21 mm. These offsets resulted in approximately ± 2 N (mean) and ± 42 N (mean $\pm$ SD) of worst-case bite force calculation error at the maximum bite location (Figure S2 and Table S2).

Figure S2: Effect of propagated kinematic measurement errors on calculations of bite force for the right 1st molar across the chewing cycle (a) and resulting mean error in the calculated bite force resulting from ± 1 SD of the kinematic measurement error (b). The black line represents

the nominal bite force, dark blue lines represents the error resulting from the measurement error of the DMD system and light blue lines represents the error resulting from one standard deviation of the measurement error of the DMD system.

Table S2: Bite force magnitude over four propagated measurement error simulations at 25%, 50%, 75% and 100% of the chewing cycle. Results are given for the nominal mandibular motion and that offset in the z-direction by the mean kinematic measurement error (± 0.01 mm), and the mean kinematic measurement error plus or minus one standard deviation (± 0.21 mm).

Cycle progression	Nominal	Nominal + 0.01 mm	Nominal - 0.01 mm	Nominal + 0.21 mm	Nominal - 0.21 mm
25%	50.6	51.2	50.1	61.6	44.1
50%	102.2	102.5	100.8	119.8	91.1
75%	178.2	179.3	176.3	204.6	163.3
100%	310.7	312.9	308.9	354.6	270.7