Effect of bidirectional loading on contact and force characteristics under a newly developed masticatory simulator with a dual-direction loading system

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Mechanical responses of the test specimen under bidirectional and unidirectional loading were investigated using a newly developed masticatory simulator. The simulator adopted a four-bar linkage mechanism to create both loading patterns. Scratch/occlusal contact characteristics, contact force profiles and the fracture of restored tooth samples were investigated. With bidirectional loading, which imitates the nature of human chewing cycle closer than the unidirectional loading does, the occlusal contact was ovoid in shape whereas a small circular area was observed from the test with unidirectional loading. The contact force profiles were also noticeably dependent on the loading patterns. Measured contact forces from bidirectional loading were more uniform than those from unidirectional loading. Bidirectional loading also induced the cuspal fracture with similar characteristics of natural cuspal fractures in humans. The differences of force characteristics between those of bidirectional and unidirectional loadings emphasize the importance of employing bidirectional loading in dental material testing.

Keywords: Chewing, Simulator, Bidirectional, Dental material, Force characteristics

INTRODUCTION

Preclinical testing of the mechanical properties of new dental materials via in vitro chewing simulation is prerequisite before a clinical trial because an initial in vivo testing is time-consuming, costly and ethical related. Laboratory simulation of human chewing is thus undoubtedly important in preclinical testing of a new dental material. Masticatory simulators have also been implicated in several preclinical tests including fracture resistance test, fatigue test and wear test1-5. In addition, the simulators are used to simulate aging before an ultimate load test6. An in vitro experiment of a dental material should closely imitate physiological characteristics of human chewing, including the direction of jaw movement7. In human, although mandibular movement is three-directional, the force direction at the tooth-to-tooth occlusal contact appears to be bidirectional8, resulting in the bidirectional, not unidirectional, force at the occlusal contact area during human chewing cycles.

The direction of force has a significant impact on the magnitude of the normal and lateral load components9. Lateral loads increase the amount of shear forces transmitted from the occluding point, on either tooth structures or dental materials, to surrounding tooth structures, periodontal ligament (PDL) and alveolar bone. Sustained loading due to masticatory force on occlusal tooth surfaces may thus result in unfavorable outcome such as broken restored dental materials and tooth or root fractures. The simulator with the most mimicking human jaw movement is therefore indispensable for in vitro dental material testings.

Most of currently available masticatory simulators, including the universal testing machine, have not been able to imitate all key parameters of human chewing especially the restricted one directional occlusal force. The limited availability of simulators that provide virtual bidirectional forces at the occluding point makes the testing results highly varied and difficult to translate into clinical situation. However, the differences of contact and force characteristics generated by bidirectional and unidirectional loadings have not yet been clearly demonstrated. It was hypothesized that when compared with unidirectional loading, bidirectional loading during chewing cycles would result in marked different characteristics associated with tooth contact, and thus influencing the results of dental material testings.

In order to unequivocally compare the effect of bidirectional and unidirectional loadings on contact and force characteristics, it is necessary to develop a simulator with an ability to mimic both loading patterns. Therefore, we first developed a masticatory simulator mimicking the natural human chewing that simulated both bidirectional and unidirectional motions of the human mandible. We then examined the effect of bidirectional loading on the contact and force characteristics when compared with the unidirectional loading under the developed simulator. We reported here that the recently developed simulator represented more complex oral masticatory stress than those currently available machines and that the contact and force characteristics of samples tested under bidirectional loading significantly differed from those tested under unidirectional loading. This emphasizes the importance of the use of bidirectional loading force for dental
MATERIALS AND METHODS

Design of the masticatory simulator

The currently developed simulator was designed to accommodate both the bidirectional path and the conventional unidirectional path commonly used in previous masticatory simulators. The schematic of the proposed masticatory simulator is presented in Fig. 1A. There are two main parts of simulator, i.e., the upper part that replicates the maxilla and the lower part that creates the motion of the mandible. A simple four-bar linkage used in a study by Xu and colleagues \(^{10}\) was adopted to create the movement of the mandible. The four-bar mechanism was mainly composed of base link \(A_B\), crank \(A_A\), follower \(B_B\), and coupler \(A_BP\). The base link \(A_B\) was an immovable link with an adjustable length to accommodate the bidirectional and unidirectional motions of the mandible. For the bidirectional motion, the length of the base link \(A_B\) was set at 50 mm. This length was adjusted to be 38 mm for the unidirectional motion. Crank \(A_A\) and follower \(B_B\) are 10 and 30 mm in length, respectively. The coupler \(A_BP\) was connected to crank \(A_A\) and follower \(B_B\) by two sets of ball bearings at point \(A\) and \(B\), respectively. Dimensions of the coupler \(A_BP\) are as follows: \(AB\) is 35-mm long, \(BP\) is 30-mm long, and angle \(ABP\) is 120°. With the specified dimensions, the motion of point \(P\) of the coupler \(A_BP\) was similar to the motion of the mandible if the simulator was set with a bidirectional-motion configuration. On the other hand, if the mechanism was set for the unidirectional motion, the motion of the lower compartment of the simulator was almost in the vertical up-and-down direction. The specimen base was rigidly fixed to point \(P\) of the coupler \(A_BP\), so that the specimen moved in the same path as point \(P\).

A high strength and high hardness tungsten carbide stylus was equipped on the upper part of the simulator. To obtain controllable impact forces between the stylus and the specimen, adjustable dead weights were placed on top of the upper part of the simulator. The whole part of the upper setup was placed on a rigid block and could be moved up and down along the guided columns. A miniature load cell was placed between the tungsten carbide stylus and the movable weight base. The load cell was employed to measure the normal force induced during the impact between the specimen and the stylus. The cranks were driven by a motor with an adjustable speed of rotation up to 2 Hz. The signals of measured forces from the load cell were transmitted to a sensor interface (PCD-300A, Kyowa, Japan) which was connected to a personal computer. The sampling frequency of the sensor interface could be set as high as 5 kHz, allowing the impact force between the stylus and the specimen being accurately measured and recorded.

The upper moveable part of the masticatory simulator was placed on the fixed base which was secured on two main columns. The vertical position of the fixed base was adjustable to accommodate various types of specimens. Most of the components were manufactured from stainless steel grade 304, except all of the shafts which were fabricated from SCM440 stainless steel. All mechanical parts of the simulator were designed based on the maximum chewing force of 500 N \(^{11-15}\) and the chewing frequency of 2 Hz. With the load capacity and chewing frequency, a 3-phase induction AC motor with a set of build-in gear box (GTR G3 series, Nissei, Japan) was selected to drive the shaft at \(A\) in the present simulator. The motor had an output speed of 1,720 rpm at a rated frequency of 60 Hz. The gear ratio of the gearbox is 1:15 with output torque of 18.6 N-m.
An inverter (J1000 series, Yaskawa, IL, USA) was used to control and convert fixed frequency single phase AC input to variable frequency three phase AC output for the motor.

**Examination of chewing path characteristics**

After the assembly, the simulator was tested and verified according to the design parameters. First, the chewing path of the test specimen, which was fixed at point P on the coupler, was measured and compared with the simulated path of the mechanism. In this part of the verification, the lower part (representing the mandible) of the simulator was operated at a relatively low speed of 0.4 Hz with the upper part removed from the setup, and the motion of the lower coupler was recorded as a video file on a digital camera (OMD-EM5, Olympus Thailand, Bangkok, Thailand). The camera was set up such that the line of sight was perpendicular to plane $\text{ABP}$ of the coupler. A number of pictures from the video file were extracted and saved as separated picture files. Each of the picture files was then analyzed using a “TechDig” software (shareware by Ronald B. Jones), which is capable of digitizing the position of a pixel in the picture. With this analysis, the location of point P on the coupler could be traced as the crank rotates in one cycle. The measured path of point P was then compared with the simulated path according to the design.

**Examination of the scratch characteristics**

A tungsten carbide stylus with a 3-mm diameter ball-ended tip (hardness of 81.5–95.5 HRA) was placed on the upper part of the simulator, while a dummy specimen was installed on the lower part. The dummy specimen was a one-inch cube of stainless steel (hardness of 55.5 HRA) with a flat and smooth surface. Since the hardness of the tungsten carbide is very much higher than that of the stainless steel, the dummy specimen was scratched after the contact. In order to avoid the effect of a curved and non-smooth tooth surface on the chewing force profile, a flat and smooth surface of the dummy specimen was used instead of the human tooth specimen in this part of the experiments. The contact force profiles were also recorded on the data acquisition system. The dummy specimen was loaded for 50 cycles with 6.8 kg of the upper movable part at the rotational frequency of 1 Hz with both bidirectional and unidirectional motions of the simulator. For bidirectional loading, the vertical tooth movement during occlusion (VTMO) was varied as 0.1, 0.3 and 0.5 mm, while the VTMO was set at 0.5 mm for unidirectional loading. Practically, VTMO is the distance of the vertical motion of the upper moveable part of the simulator. The scratches and the length of scratches were photographed and measured using a digital microscope (Dino-Lite, AnMo Electronics, Taiwan) and capturing software (DinoCapture 2.0, AnMo Electronics).

**Collection and preparation of tooth samples**

Healthy permanent maxillary premolars extracted for orthodontic proposes were collected and stored in 0.2% thymol for less than 3 months. The ethical consideration for the use of the teeth was approved by Research Ethical Committee, Thammasat University, Thailand. Each tooth was cleaned and examined using a fiber optic light under stereomicroscope and teeth with cracks or other visible defects were excluded. All samples were mounted in self-curing acrylic (Lang Dental Manufacturing, IL, USA) to 1.5 mm below the CEJ, which is approximately the level of the alveolar bone in a healthy tooth. The samples were used for the assessment of occlusal tooth contact characteristics and fracture test, as described below.

**Occlusal tooth contact characteristics**

Occlusal tooth contact has been shown to influence occlusal force distribution and this appeared to be an ‘area’, rather than a ‘point’. Thus, the patterns of tooth-stylus contact following bidirectional and unidirectional loadings were also determined. Samples of premolars were placed on the lower compartment of the simulator and subsequently subjected to bidirectional and unidirectional loadings of approximately 100 N at the rotating frequency of 1 Hz. The occlusal contact pattern was examined using red/blue dental articulating papers with a 0.0025 inches thickness (Homedent Group, Bangkok, Thailand). The articulating paper masks were visualized and the photos were taken using a digital camera (OMD-EM5).

**Contact force measurement**

In order to determine the influence of bidirectional loading on the contact force at various applied dead weights and chewing frequencies, two sets of experiments were performed, i.e., the experiments with varying dead weight (fixed frequency) and the experiments with varying frequency (fixed dead weight) under both bidirectional and unidirectional loadings. In the first part, the setup was operated at a fixed rotation frequency of 1 Hz with the dead weight varied from 6.8 to 14.8 kg (corresponding to a static compressive force of 66.7 to 145.2 N, respectively). The other set of experiments was conducted using a fixed applied load of 9.8 kg. The rotation frequencies were varied from 0.1 to 1.9 Hz. Contact forces were recorded on the data acquisition system. Both bidirectional and unidirectional motions of the lower part of the simulator were performed for each of the testing conditions. The measured force from the load cell was recorded at the sampling rate of 5 kHz. The average of measured forces during a number of contacts between the stylus and the dummy specimen was reported as the average contact force. The maximum value of the measured contact force was also reported as the maximum contact force.

**In vitro tooth fracture test**

In order to determine the influence of bidirectional and unidirectional loadings on the fracture of tooth samples, human teeth with restorations were subjected to both occlusal loading patterns, and the number of cycles to
failure and the failure characteristic were examined. Maxillary premolars were randomly divided into 2 groups (n=5 each), i.e., Group 1 bidirectional loading and Group 2 unidirectional loading. The mesio-occluso-distal (MOD) cavity was prepared. The thickness of the remaining tooth surface was approximately 2 mm and the gingival wall was finished at the level of 0.5 mm occlusally to the cement-enamel junction (CEJ). A new bur was employed for every 2 teeth. The thickness of the cavity walls was standardized with an orthometer gauge. All samples were restored with a resin-based composite (Filtek™ P60 Posterior Restorative, 3M ESPA, Seefeld, Germany) and an adhesive system (CLEARFIL™ SE BOND, Kuraray America, USA). The restored tooth specimens were then set up on the lower part of the simulator while the tungsten carbide stylus was mounted on the upper part of the setup. The area of applied contact force was set at the buccal incline plane of the palatal cusp.

To optimize the time of the experiment and obtain reasonably uniform force profiles, the rotational frequency was selected to be 1 Hz for all experiments in this part. The applied dead weight on the upper part of the simulator was chosen to be 19.8 kg (approximately 200 N). The force profiles and the number of loading cycle were recorded on the data acquisition system. All 10 specimens were tested until the fracture was observed or 100,000 loading cycles were reached. It was assumed that the specimen became fractured if the measured contact force was suddenly dropped. Pictures of the specimens before and after the tests were photographed using a digital camera (OMD-EM5).

RESULTS

Operation of the masticatory simulator

The designed simulator was manufactured and assembled as shown in Fig. 1B. A pair of tooth specimens was firmly fixed on the upper and lower part of the simulator to demonstrate the contact configuration between the upper and lower teeth. It was clearly seen that the lower specimen moved in both vertical and horizontal directions for the testing with bidirectional loading. In the default setting, when a tooth, which was previously fixed on the specimen holder of the lower part, moved up and reached the other tooth, the whole set of weight base, applied dead weight and the upper specimen moved up along the guided columns for a distance of approximately 0.5 mm. At this moment, the specimen was subjected to a compressive load due to weight of the upper movable part and the additional dead weight. The compressive load was observed, and the corresponding contact forces were subsequently recorded by the data acquisition system. The applied load decreased to zero when the specimen moved down and the upper set of weight moved to rest on the rigid block. This completed a chewing cycle, which is corresponding to one cycle of rotation of crank AAo.

Chewing paths

In Fig. 1C, the results show that the simulated paths were continuous and smooth whereas the measured paths, especially for the bidirectional loading, were somewhat not as smooth as the simulated one. Imperfection of the actual paths was probably attributed to the unavoidable influence of the tolerance and precision of all connecting parts, for example pins and bearing. It is important to note that during a chewing cycle, the stylus contacted the specimen only when the specimen moved to the top region of the chewing path, which was the only part in the chewing path influencing the contact force characteristics. The results in Fig. 1C demonstrate that the top region (arrow heads) of the actual chewing path well corresponded with the designed path for both bidirectional and unidirectional motions. The top region of the bidirectional loading path clearly consisted of both lateral and vertical motions whereas that of the unidirectional loading path possessed mainly only the vertical motion (Fig. 1C).

Characteristics of scratches on the dummy and occlusal contact marks

We first compared the contact mark patterns during bidirectional and unidirectional loadings of the simulator by examining the scratches (contact marks) on the dummy after the stylus-dummy contact. The results in Fig. 2A show that the scratches on the dummy surface for both bidirectional and unidirectional loadings were completely different. For bidirectional loading, the scratches appeared as a line with an approximate length of 3.6, 2.9 and 1.3 mm, for the test with VTMO=0.5, 0.3 and 0.1 mm, respectively. In contrast, a small circular scratch with a diameter of less than 0.8 mm was observed from the test with unidirectional loading (Fig. 2A). The stylus-tooth contact was also examined under both bidirectional and unidirectional loadings, and the representative articulating paper marks on the occlusal surface are shown in Figs. 2B and C, respectively. With bidirectional loading, the occlusal contact area observed on the premolar was ovoid in shape (Fig. 2B) whereas the occlusal contact mark derived from unidirectional loading appeared to be only a small circle (Fig. 2C).

Effect of bidirectional loading on the force profile at various applied dead weights

Plots of the measured stylus-dummy contact force versus time in millisecond (ms) are shown in Figs. 3A–C and Figs. 3D–F for bidirectional and unidirectional loadings, respectively. For both loading patterns, each stylus-dummy contact in a cycle took approximately 100 ms. It was clearly seen that the contact force profiles were markedly dependent on the loading configuration. Generally, the contact force was highly fluctuated in the first 20 ms of the contact for both bidirectional and unidirectional loadings, representing the first phase of the contact. The next 70 ms following the first phase formed the second phase of the contact. In this phase, the degree of fluctuation of the contact force was decreased compared with that of the first phase of the contact. The
second phase of the contact force profile was significantly influenced by the loading patterns. For bidirectional loading, the contact force in the second phase was almost constant in the experiments with low applied dead weights (Fig. 3A). The variation of contact force over time was higher when the applied dead weight was increased, resulting in 4 distinct fluctuation peaks being observed in the second phase of the contact cycle for the experiments with applied dead weights of 10.8–14.8 kg, as shown in Figs. 3B and C. On the other hand, only 2 obvious peaks were observed from the experiment with unidirectional loading (Figs. 3D–F). In the third phase of the contact, the measured contact force was fairly oscillated, similar to the first phase of the contact. The third phase of contact profile from bidirectional loading appeared to be similar to that of from the unidirectional loading.

In view of contact mechanics, it is typical that the measured force is fluctuated during the contact. Thus, it is reasonable to use the average value of the measurements to represent the contact force. The average and maximum contact forces in percentage comparing with applied dead weights between the stylus and the dummy specimen at different applied dead weights, obtained from three measurements, is presented in Fig. 4A. Theoretically, if the test is performed with a very low rotational frequency, the contact force in the second phase should be constant and well corresponds to the applied dead weight. However, by using a physiologic human chewing frequency of 1 Hz, the average measured contact force in case of the bidirectional loading was 6–14% higher than the applied dead weight while the average measured contact force under the unidirectional loading was about 1–5% lower than the applied dead weight (Fig. 4A). On the other hand, the maximum measured forces in the case of unidirectional loading were generally higher than those with the bidirectional loading. The results in Fig. 4A also show that, regardless of the dead weight used, the measured contact forces from bidirectional loading seemed to be less fluctuated than those from unidirectional loading, as shown by smaller differences between maximum and average contact forces.

**Effect of bidirectional loading on the force profile at various chewing frequencies**

The force profiles obtained from the simulator operating under bidirectional and unidirectional loadings with various rotational frequencies and a constant dead weight of 9.8 kg (96.4 N) were presented in Fig. 5. The force profiles from the tests with a rotational frequency of less than 1 Hz can be divided into 3 phases similar to the profiles in Fig. 3. On the contrary, the force profiles of the tests with a rotational frequency of higher than 1 Hz did not exhibit an apparent constant force in the middle phase of the profiles. The results with lower rotational frequency show the fluctuation of the measured contact forces in the first and third phases of contact regardless of the loading type. However, the type of loading influenced the deviation of measured contact force in the second phase of the force profile (Figs 5A–C vs. Figs. 5D–F). At rotational frequencies between 0.1 and 0.4 Hz, the majority of the measured contact forces in the second phase appeared to be constant under bidirectional loading, while they slightly varied under unidirectional loading, as shown in Fig. 5A vs. Fig. 5D. An increasing degree of fluctuation of the measured contact forces was noticed as the rotational frequency increased, and with the rotational frequency higher than 1 Hz, no region of constant force was clearly observed (Figs. 5A–C and Figs. 5D–F). Again, at all tested frequencies higher than 0.4 Hz, higher degree of fluctuation of the contact forces seemed to be observed in unidirectional loading compared with the other type of loading.
The measurements were performed at the rotational frequency of 1 Hz. and different applied dead weights (A), and at the applied dead weight of 9.8 kg with different rotational frequencies (B).

The average and maximum contact forces in percentage comparing with applied dead weights between the stylus and the dummy specimen at different rotational frequencies for both types of loading obtained from three measurements is presented in Fig. 4B. The results revealed that the maximum force was increased with the rotational frequency for both bidirectional and unidirectional loadings, but the average force did not obviously vary with the rotational frequency. Similar to the previous set of experiments with varying the applied dead weight, the average contact force during the bidirectional loading was higher than that during the unidirectional loading.
the unidirectional loading regardless of the rotational frequency. For bidirectional loading, the average force was almost independent of the rotational frequency, i.e., varied in the range of 105 to 110% of applied dead weight with the rotational frequencies between 0.1 and 1.8 Hz. The average force dropped to 99.6% of applied dead weight with the rotational frequency of 1.9 Hz. For unidirectional loading, the average force varied between 99 to 102% of applied dead weight with the rotational frequencies of ≤1 Hz. For higher rotational frequencies of more than 1 Hz, the average force trended to decrease with increasing rotational frequencies. In addition, the results in Fig. 4B demonstrate that regardless of rotational frequency, the measured contact forces from bidirectional loading seemed to be less fluctuated (more uniform) than those from unidirectional loading, as shown by smaller differences between maximum and average contact forces.

Effect of bidirectional loading on in vitro tooth fracture

Using an applied dead weight of approximately 200 N, all five specimens under bidirectional loading were fractured after 72, 241, 577, 2,073 and 5,081 loading cycles. The average total number of cycles to failure was approximately 1,609 cycles. A representative photo in Fig. 6A presents a typical characteristic of a specimen following the fracture test. The results showed the fracture of palatal cusp together with the failure at the tooth-restoration interface, and the fracture extended to the cervical third portion near the area of CEJ (Fig. 6A). On the other hand, following 100,000 loading cycles under unidirectional loading, all five specimens were not catastrophically destructive. Only chipped enamel was

Fig. 5 Contact force profiles under the bidirectional loading (A–C) and the unidirectional loading (D–F) with various frequencies.

Fig. 6 Fracture tests under bidirectional and unidirectional loadings of the masticatory simulator.
MOD restored premolars were used as described in the Materials and Methods. A representative fractured sample under the bidirectional loading is shown in (A), and the sample after the 100,000 loading cycles under unidirectional loading is shown in (B). The representative force profiles from the tests under both bidirectional (C) and unidirectional (D) loadings are presented.
observed at the tooth-stylus contact spot on the occlusal surface, as shown in Fig. 6B.

Representative force profiles acquired from the fracture tests under both bidirectional and unidirectional loadings are shown in Figs. 6C and D, respectively. Under bidirectional loading, the force profile of a specimen which was fractured after 241 loading cycles is presented in Fig. 6C. The force profile was oscillated similar to a sinusoidal wave in the first 60 ms or first half of the cycle, approximately. The force profile in the interval of about 30 ms in the second half of the cycle seems to be constant. The force profile from the test with unidirectional loading is shown in Fig. 6D. It is noticed that the trend of the loading profile in this test is similar to a sine wave with decreasing amplitude in the first 90 ms. No constant measured force was observed throughout the loading cycle.

**DISCUSSION**

Testing of dental materials using a human mimicking masticatory simulator is of importance in preclinical study before a clinical trial. Experiments using dynamic loading appear more clinically relevant. Although a human mandible moves in three-directional fashion, the force at the occlusal contact surface appears to be bidirectional. However, most of currently available masticatory simulators, including the universal testing machine, have not been able to generate mimicking chewing force and direction applied on a tooth specimen during dynamic loading. Most of the available simulators are limited to only unidirectional loading configuration. This limitation could possibly make the test results markedly vary and unable to translate to clinical situation. Using the presently developed simulator, the results showed that key characteristics of the contact and force applied on the tested samples under bidirectional loading significantly differed from those under unidirectional loading. This emphasizes the importance of the use of the bidirectional, not unidirectional, loading system for dental material testings. Moreover, the proposed masticatory simulator is capable of simulate scratches with a variety of lengths by controlling the VTMO.

The masticatory simulator in the present study was designed to simulate both bidirectional and unidirectional motions of the mandible. Naturally, the mandible moves in vertical and horizontal directions to form a bidirectional path during chewing, and bidirectional force is generally generated at the occluding point. As a result, teeth are subjected to both normal and shear forces during tooth-to-tooth contact. On the other hand, during unidirectional motion, the mandible moves up and down in almost only vertical direction, so that the majority of the force generated during the tooth contact is a normal force with only a minimal amount of shear force. From the measured path of the lower compartment, the currently developed simulator was successfully designed to move biaxially, thus theoretically generating both shear and normal forces during the contact. It is important to note that the standard deviations of the average measured contact force appeared to be low for all cases of the test. This observation suggests that the custom-made simulator with the use of dead weight and the average force to represent the force applied to the specimen is reasonably acceptable.

With the bidirectional loading configuration, the contact marks from the tooth-stylus testing appear as an oval area which corresponded very well with the contact marks from the stylus-dummy testing. The oval contact area on the tooth specimen with the bidirectional-loading configuration closely resembles the contact area on the occlusal surface of human posterior teeth. Since occlusal tooth contact has been shown to influence occlusal force distribution, it is likely that the distribution of occlusal force within the tooth sample tested with bidirectional loading is different from that tested under unidirectional loading. Therefore, it is possible that compared with unidirectional loading, this *in vitro* bidirectional loading may provide the force distribution, from the occlusal contact area, that resembles the *in vivo* intra-oral situation.

The present results demonstrated that independent of the dead weight and rotational frequency, tests with bidirectional loading possessed more homogeneous contact forces than those derived from unidirectional loading, as shown by smaller differences between maximum and average contact forces (Fig. 4). Because most chewing simulators used in previous studies operated with a unidirectional loading fashion, this may result in a nonuniform force loaded on the tested samples, thus causing a high standard deviation, a poor reproducibility and a large variation among previously published results. It is thus encouraged that bidirectional loading with standardized masticatory testing parameters should be used in testing dental materials.

From the results in Fig. 4A and most of Fig. 4B, the average contact force from the bidirectional testing was higher than the applied dead weight, but was lower in case of the unidirectional loading. The higher contact force in case of the bidirectional motion was probably attributed to the friction between the upper movable part and the guided columns. For the test with bidirectional motion, the sample collided with the stylus in an inclined direction, resulting in a vertical movement of the upper movable part and a significant magnitude of a normal force between the upper movable part and all four guided columns. The normal force between two surfaces produced friction force in the opposite direction of the motion. This normal force was not significantly presented in the unidirectional testing. Therefore, in addition to the force from the dead weight, there was a friction force applied to the upper movable part applied in the downward direction. This friction force was probably the cause of the higher average force in the case of the bidirectional motion. On the contrary, most of the measured contact forces from the unidirectional test were lower that the applied dead weight. These lower
measured loads were probably the result of multiple impacts between the stylus and the dummy. In Figs. 3A–C, most of the force profiles of the bidirectional motion were smooth whereas the force profiles of the unidirectional motion were fluctuated, as shown in Figs. 3D–F. It implied that, with the unidirectional motion, multiple impacts between the stylus and the specimen were presented throughout the cycle of contact. The presence of multiple impact indicated that the contact between two surfaces was not perfect, thus the force from the dead weight was not perfectly applied to the specimen. As a result, the measured contact force was lower than expected. The present results suggest that the direction of the impact force is one of the factors that control the magnitude and degree of fluctuation of the contact force. It is noteworthy that the motion of most of the commercially available chewing simulators appears to be similar to the unidirectional motion, thus possibly producing non-uniform force profile. Besides using spring-damper system, low testing velocity and low-stiffness specimen or stylus, the motion characteristics of the specimen or counter-specimen could also be considered in order to obtain the most uniform contact force.

Cuspal fracture is a natural and deteriorating consequence of tooth weakness and/or parafunctional force, and the study of this phenomenon is thus clinically relevant because it is widely observed in dental practice\(^{20-23}\). The incidence of cuspal fracture has been reported to be frequent in upper premolars\(^{24,25}\). Moreover, the lingual cusp of posterior teeth fractures more often than the buccal cusp does, and the fractures may extend to a level either above the gingival crest, at the crest or below the crest\(^{26}\). In the present study, the simulator was used to induce cuspal fracture of MOD resin composite-restored maxillary premolars under both bidirectional and unidirectional loadings. The results showed that the simulator operating with the bidirectional loading was able to simulate the occurrence of cuspal fracture with similar characteristics of natural cuspal fractures in humans. It is noteworthy that characteristics of force applied on tooth samples produced by bidirectional, but not unidirectional, loading may be comparable to those generated by complex intra-oral masticatory forces. From the force profiles in Figs. 6C and D, it is possible that force distribution in the restored tooth samples generated by bidirectional and unidirectional loading are markedly different. The force profile from the bidirectional loading appeared to be more uniform than that of the unidirectional loading. It can be hypothesized that the multiple impacts in the test with unidirectional loading caused the lower-than-expected average load applied to the specimen. Further studies using computational-based modeling, such as finite element analysis, are required to investigate this hypothesis. The present results of the cuspal fracture test using the custom-made masticatory simulator also suggest that the simulator with the capacity to generate human mimicking masticatory bidirectional force may be used as a useful tool to investigate the fracture resistance of new materials and teeth restored with new techniques.

In addition to type of force configuration investigated in the present study, other key parameters also play an important part on the force characteristics occurred on the tested samples. Because the present custom-made masticatory simulator was also designed to possess a range of adjustable features, including magnitude of loading/chewing force and chewing frequency, the role of a wide range of clinically relevant parameters in the force characteristics on new dental materials can be tested under the most mimicking condition of complex intra-oral masticatory forces. For example, it is possible to determine the effect of forces, from physiologic chewing forces of 10–120 N\(^{18,26-28}\) to more excessive parafunctional bite forces associated with bruxism and clenching of approximately 200–800 N\(^{16,29}\), on fracture resistance of a material of interest. However, a number of other factors, such as moisture, temperature and pH, also influence the mechanical properties and behavior of materials in the oral cavity\(^{30,31}\). We are currently incorporating accessories that add the aforementioned factors into the test condition while using the simulator. This will help the user extend the use of machine for a wider variety of tests and investigate the factor affecting new dental materials in the most mimicking intra-oral environment of human.

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