Three-dimensional orthodontic force measurements

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Introduction: Until recently, much of the orthodontic biomechanics literature was restricted to 2-dimensional experimental studies and, more recently, to assumption-based 3-dimensional computer modeling. There is little evidence in the literature regarding 3-dimensional experimental measurements and analysis of orthodontic force systems. Methods: The purpose of this study was the design, construction, and validation of a laboratory-based human mouth model capable of accurately measuring forces and moments applied by orthodontic fixed appliances on all teeth in 1 arch. A high canine malocclusion was simulated, and forces and moments acting on the canine, lateral incisor, and premolar were measured with passive and conventional ligation. Results: We were successful in building this human mouth model. The error in force measurements of the 14 transducers was 1.54%. The force system resulting from passive ligation brackets was considerably different from that of conventional ligation. Conclusions: This method will allow us, for the first time in the history of our specialty, to determine with great accuracy the forces acting on orthodontically treated teeth. Future research will focus on simulating many types of orthodontic clinical applications of full-fixed or partial-fixed appliances. (Am J Orthod Dentofacial Orthop 2009;136:518-28)

To understand orthodontic tooth movement, clinicians and researchers must first study its biophysics to define and quantify the force systems applied to the teeth before making valid judgments about the clinical responses of those teeth to orthodontic forces. Many undesirable tooth movements during orthodontic treatment can be attributed directly or indirectly to the lack of understanding of the physics of tooth movement. Many variables influencing orthodontic treatment cannot be fully controlled, such as growth and tissue response to appliances. However, the force placed on the tooth should be a controllable variable, and careful study of the physics underlying our clinical applications can help in reducing undesirable side effects.1

A tooth’s response to a force can be studied at 3 levels: clinical, cellular, and biomechanical. The clinical level allows the study of phenomena such as rate and direction of tooth movement, pain response, and tooth mobility. The cellular level gives insight into the dynamics of the biology of tooth movement, including the dynamics of bone and connective tissue. The vaguely understood biomechanical level provides the ability to accurately determine the level of stress in various areas of the periodontal ligament (PDL); this might be the best means of correlating the application of a force on a tooth with the tooth’s response.2,3

Until recently, much of the orthodontic biomechanics literature was restricted to 2-dimensional experimental studies of the biomechanical aspects of orthodontic force systems and, more recently, to 3-dimensional (3D) computer modeling. There is little evidence regarding 3D experimental measurements and analysis of orthodontic force systems.4-11 To study the orthodontic force system in 3 dimensions, one would need a force sensor capable of measuring 3D forces and moments. Furthermore, those measurements must be made on all teeth in a dental arch simultaneously. The sensors therefore must be small, a size that can be used on all teeth in a dental arch. Such a small 3D sensor does not yet exist. Three-dimensional force and moment measurement technology is commercially available but not in tooth-size...
dimensions. These 3D sensors are called multi-axis force transducers. Using such sensors in orthodontic research requires complicated engineering designs, micromachined mechanical parts, and specialized software. Lapatki et al \textsuperscript{9} and Lapatki and Paul \textsuperscript{12} at the University of Freiburg in Germany designed and tested a promising experimental bracket-sized 3D sensor, but more work is needed on their design to permit more accurate measurement of forces higher than 1.5 N (approximately 150 g).

The purpose of this article was to report on the design and construction of a device that can show the details of the force systems used in modern orthodontic mechano-therapy. Our aim was to use these multi-axis force and moment transducers to help to simultaneously quantify the forces applied to all teeth in 1 arch in real time when orthodontic fixed appliances are used. Many orthodontic clinical applications can be studied by using this device.

\textbf{MATERIAL AND METHODS}

The orthodontic simulator (OSIM, Figs 1 and 2) was built and tested through collaborative work between the Division of Orthodontics and the Department of Mechanical Engineering at the University of Alberta, Edmonton, Alberta, Canada. The OSIM is essentially a human mouth model with 1 dental arch containing 14 teeth, to which brackets can be bonded and wires ligated. This model can simultaneously measure the force systems applied on all teeth in the arch by orthodontic appliances. The software application performs the mathematical calculations and generates real-time 3D displays of forces acting on every tooth simultaneously. Below is a detailed description of this apparatus.

The hardware includes force and torque transducers. A load cell or transducer is a mechanical device that can measure the forces and moments (torques) applied to the top of the load cell. Industrial Automation Nano17 load cells (ATI Industrial Automation, Apex, NC) (Fig 3) were used to measure the 3 force and 3 moment components of the applied loads. The compact transducer is the smallest commercially available 3D load cell that uses silicon strain gauges to sense forces. Each load...
Table I. Maximum full-scale measurement uncertainties for the Nano17 transducer (error) (provided by the manufacturer)

<table>
<thead>
<tr>
<th>Calibration</th>
<th>Fx</th>
<th>Fy</th>
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<th>Tx</th>
<th>Ty</th>
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<tr>
<td>Nano17</td>
<td>SI-59-0.5</td>
<td>1.50%</td>
<td>1.50%</td>
<td>2.00%</td>
<td>1.50%</td>
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<tr>
<td>Resolution DAQ card (N)</td>
<td>1/1280</td>
<td>1/1280</td>
<td>1/1280</td>
<td>1/256</td>
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*Nmm, Newton millimeter; DAQ, data acquisition.*

The OSIM device (Figs 1 and 2) provides the means of connecting each tooth in a dental arch to a multi-axis force transducer. To build a dental arch containing 14 teeth, and connect each tooth to a 3D load cell 17 mm in diameter, we had to design a special connector that incorporated vertical (MTI-153203, MIC 0-25 MM head, Mitutoyo, Kawasaki, Japan; Fig 2, a) and horizontal (precision micrometer drive with nonrotating tip, 10 mm travel, M-631.00, Physik Instrumente, Karlsruhe, Germany; Fig 2, b) nonrotating micrometer heads (with nonrotating spindles). Turning the horizontal micrometer head produced horizontal tooth movement, and turning the vertical micrometer head produced vertical tooth movement; the resolution of those movements was 0.01 mm. Complete engineering drawings (Fig 1) were made. The major components were made of aluminum because of its light weight, stiffness, and ease of machining. Smaller components were made of stainless steel and brass.

The model contains a base plate that holds 14 vertical micrometers with the nonrotating spindles. The load cells were mounted on top of the vertical micrometer spindles. Another part, mounted on top of the load cell, holds the horizontal micrometer. The spindle of the horizontal micrometer was replaced with a custom-made attachment that holds a platform on which artificial teeth with brackets were mounted.

Three types of artificial teeth were prepared: (1) aluminum cylinders 5 mm in diameter with a flat surface to allow accurate orientation, with no interproximal contacts; (2) aluminum cylinders with diameters equal to the mesiodistal widths of teeth, to provide interproximal contacts between the cylinders; and (3) anatomically correct teeth machined of aluminum in a multi-axis computer numerical controlled machine.

Epoxy resin was used to bond metal brackets to the aluminum cylinders, by using a jig to guarantee that all brackets were mounted at 0° tip and torque.

The sensors of the coordinate measurement machine were connected to the teeth via a custom-made connector, which meant that the loads were measured at a point other than the point of application. Therefore, force system transformations called Jacobian transformations were necessary to transform the force system from the load cell frame to the tooth frame; this process consisted of a number of matrix multiplications. To perform those transformations, we needed to know...
accurately the x, y, and z coordinates and the orientations of the tooth in relation to its designated transducer. We used a coordinate measurement machine (platinum 4 foot, FaroArm, Faro Technologies, Lake Mary, Fla) to accurately measure the position of each tooth relative to its corresponding load cell; then these positional data were transferred to custom software (MATLAB computer programming, MathWorks, Natick, Mass) for the force system transformations. Those mathematical transformations were performed in real time as data were gathered from the 14 transducers.

Two software packages were used: third-party data acquisition software (LabView, National Instruments, Austin, Tex) and custom-made visualization software (OSIM). The data acquisition program was designed to perform several operations; after receiving raw voltage readings, calibrations were performed. Then the LabView software combined all readings into 1 array and sent those data to the interface between the LabView and the OSIM.

The OSIM software package was written to help interpret the data from the 14 sensors of the OSIM model, since it was necessary to provide a graphic display of the forces and moments for each load cell and tooth (Figs 4-6). It was written by using the MATLAB programming environment and provides functionality to gather and display data. The forces and moments were represented by 2 scaled arrows, one for the force and the other for the moment projected on the 3D images of the teeth (Figs 5 and 6). The sign conventions used in this software are shown in Table II.

The OSIM software package provides several functions; it can (1) perform error analysis, for checking the accuracy and precision of each transducer; (2) perform calculations on the transducer readings to transform them to the readings on the teeth; (3) show a real-time 3D graphic representation of the teeth, with the forces and moments acting on them; (4) manipulate how information is displayed graphically, by using 3D reorientation tools and varying color indexing; (5) record and sample data from the OSIM model; (6) perform statistical operations on the data; and (7) export data to an Excel, video, or image file.

The overall error in this device is from the following sources: (1) errors inherent in the transducer measurements; (2) errors from the force system transformations, which magnify the error of the load cell and are generally proportional to the moment arm (the distance between the transducer and the point of application); and (3) errors from inaccuracies in performing the transformations due to metrology error when using the coordinate measurement machine.

A calibration arm was built to measure the errors of the transducers (Fig 7). Using the calibration arm, we could apply known loads with calibration weights. The loads were applied on 1 axis at a time, by reorienting the calibration arm and using gravity to apply the loads. To ascertain that the loads were applied on only 1 axis at a time, a precision master level was used to

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<th>Table II. Sign conventions</th>
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<tr>
<td><strong>Sign</strong></td>
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<tr>
<td>Fx</td>
</tr>
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<td>Ny</td>
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make sure that the calibration arm was level and properly oriented. This meant that when a load was applied in Fx, the transducer should theoretically give zero readings on all components except Fx. If Fy was loaded, the transducer should theoretically give zero readings on all components except Fy.

Error data were collected for each of the 14 transducers by using a set of weights to apply the loads in increments of 50 g starting at zero and ending at 500 g. The calibration arm was reoriented to 6 positions to allow us to apply loads in $F_x$, $F_y$, $F_z$, $M_x$, $M_y$, and $M_z$. Two hundred samples were gathered for each weight applied, and averages were calculated.

The error is reported as a percentage by calculating the difference between actual and expected, and dividing it by the expected. For example, to report the error in Fy when the x-axis is loaded, we calculated the average error in Fy when the load was applied in the x-axis and reported it as a percentage. The same was done for all 6 components (Fx, Fy, Fz, Mx, My, and Mz). The force error data are reported in Table III, and the moment error data are reported in Table IV.

When Fx was being loaded, the overall average error was 1.84%. In Fy loading, the overall average error was 1.38%. When Fz was loaded, the overall average error was 1.75%. The overall average error in the force reading from the 14 transducers was 1.66%.

When Fx was loaded, the overall average error in moments was 2.4%. In Fy loading, the overall average error in moments was 1.53%. In Fz loading, the overall average error in moments was 2.59%. The overall average error in the moment reading from the 14 transducers was 2.17%.

The OSIM was set up to simulate a malocclusion with a high maxillary right canine (Fig 8). Indirect bracket placement by using a full-sized archwire tied to all brackets was used to ensure that all brackets were in the zero default position (zero torque, zero tip, and perfect alignment of all teeth). Damon maxillary 0.022-in standard prescription brackets (Ormco, Orange, Calif) and 0.018-in copper-nickel-titanium wires (Ormco) were used. The wire was ligated with all brackets in the start (zero) position. The maxillary right canine was moved 4 mm apically and then back to the zero position in 0.1-mm increments. This allowed us to gather the force and moment data during the loading and unloading of the wire. The same experiment was...
repeated 5 times with elastomeric ties and 5 times with passive ligation. Each time, a new wire was ligated. The objective of this experiment was to compare the force system from passive self-ligation as opposed to conventional elastic ligation in a high canine situation.

Three-dimensional force and moment data were collected for the canine, lateral incisor, and premolar for every 0.1 mm of canine movement. The temperature of the OSIM was controlled at 35° to 36°C throughout the experiment.

RESULTS

The force and moment data for the canine, lateral incisor, and premolar are shown for passive self-ligation and conventional ligation in Figures 9 and 10.

Of the vertical forces, the extrusive force (Fz) acting on the canine in passive ligation has 2 reciprocal intrusion forces on the premolar and the lateral incisor. The force on the canine follows the characteristic curve of a superelastic nickel-titanium (NiTi) alloy. During loading, the extrusive force on the canine rose to 4.98 N at 4 mm of displacement. During unloading, almost immediately, the extrusive force dropped to 2 N at 3.4 mm and remained relatively constant from 3.4 to 1.1 mm of displacement. The intrusive forces on the lateral incisor and the first premolar showed the same characteristic load-deflection curve. On the other hand, the extrusive force on the canine in the conventional ligation method followed a more linear loading curve, rising to 7.68 N at 4 mm of displacement. Then during unloading, the force dropped to 1.16 N at 2.4 mm of displacement, rose to 1.62 at 1.3 mm, and then declined again to 0 at 0.3 mm. The OSIM display in Figures 4 and 5 shows the difference in Fz forces between conventional and passive ligation after archwire loading. It is apparent that the force acting on the canine was purely vertical with passive ligation, whereas other unwanted force components (Fx and Fy) acted on the canine with conventional ligation (Figs 4 and 5).

Fx is the mesiodistal force, which in this experiment constituted friction and binding (resistance to sliding). In passive ligation, there is relatively low resistance to sliding during loading and unloading compared with conventional ligation. In conventional ligation, the canine experiences the highest resistance to sliding—2.45 N in a mesial direction at 3.9 mm during loading. In passive and conventional ligation, the lateral incisor and the first premolar experience higher resistance to sliding during unloading compared with loading. The OSIM display in Figure 5 shows the difference in mesiodistal forces between the 2 ligation methods. The mesial and distal forces on the lateral incisor and the premolar, respectively, were higher for conventional than for passive ligation.

Fy is the labiolingual force. The canine in both ligation methods started with a labial force during the initial stage of loading; this changed to a lingually directed force halfway during loading and then changed back to a labially directed force as soon as unloading began. The force magnitude acting on the canine was higher in conventional ligation compared with passive ligation. The lateral incisor experienced a lingual force during most of the loading phase; this became a labial force during unloading. The force magnitude was again much higher in the conventional ligation compared with passive ligation. The premolar showed low labial forces acting on it in the passive ligation experiment, but force magnitude was much higher in conventional ligation while maintaining the same pattern as passive ligation during loading and unloading.

The moments recorded in the conventional ligation experiment were invariably higher than those in the passive ligation experiment. The My moment is the tendency of a tooth to rotate around the buccolingual axis (tip). My is predictably tending to tip the lateral incisor distally and the premolar mesially, and it is higher in conventional than in passive ligation (Fig 6). The Mz is the rotation around the long axis of the tooth and is higher in conventional than in passive ligation. Mx is the tendency to rotate around the mesiodistal axis (torque) and is considerably higher in conventional than passive ligation.

DISCUSSION

Orthodontic knowledge in clinical biomechanics has traditionally been derived from studies that can be categorized in 2 major groups. The first is mechanical testing of orthodontic appliances, with a few mechanical testing techniques to study specific aspects of orthodontic biomechanics. The second is computer modeling and finite element analysis.

Friction has been studied in a number of ways. In some instances, the wires were pulled through at least 1 bracket,13-21 and, in other instances, a bracket was slid on a wire.22-26

Orthodontic wires have been studied in various ways.27-30 Three-point bending still is the universally accepted test for orthodontic wires, even though it does not incorporate an orthodontic bracket or a ligation method into the test and is performed on a straight section of the wire. Laboratory tests are usually simplified and designed to look at only 1 or 2 variables related to the wires tested, and that makes generalization difficult. The
bracket-wire interface varies significantly according to the ligation mechanism used. Elastomeric ties, stainless steel ligature ties, and active and passive self-ligation brackets are likely to produce different force systems. Interbracket distance is another variable often not accounted for when discussing wire properties.

The use of various designs of orthodontic loops and bypass arches is probably the best-studied aspect of orthodontic force systems.\(^5\)\(^,\)\(^31\) The reason is the large interbracket distance between a canine and a molar; measuring the forces in 3D was performed on either side of the orthodontic loops. However, orthodontic loops are taking a smaller part of contemporary orthodontic practices.

The second area of orthodontic biomechanics research is computer modeling and the use of finite element analysis.\(^32\)\(^-\)\(^36\) The fast development of computer technologies made it possible for mechanical tests to be simulated instead of performing the actual tests.\(^32\)\(^-\)\(^36\) However, to accurately perform such simulations, a detailed understanding of the components of the tested object is necessary, and equally important is how the different components interact. The most sophisticated computer modeling and simulation is by definition an

Fig 9. Forces $F_x$, $F_y$, and $F_z$ on the canine, lateral incisor, and premolar with passive and elastic ligation, with 0.018-in NiTi wire in a high canine malocclusion.
approximation to reality, and keeping the assumptions in a model to a minimum improves its predictions and simulations. How the bracket and the wire interact in various clinical applications is largely unknown, especially when we consider a full-arch fixed-appliance application. Orthodontic biomechanical force systems with more than 2 points of applications are considered indeterminate; in other words, they cannot be theoretically analyzed or directly solved from force balance. We need to measure forces in 3 dimensions in as many clinical situations as possible to build enough knowledge and then use this biomechanical data base to construct a computer model based on actual experimental studies.

We now have a device that can replicate a specific malocclusion and produce detailed 3D information regarding orthodontic forces applied on all teeth in a dental arch simultaneously. Clinicians are not generally interested in knowing the coefficient of friction for a specific type of wire when used with a specific bracket or how much of the resistance to sliding results from friction vs binding. Clinicians need to know the forces applied on the dentition when specific combinations of bracket, wire, and ligation method are used in a malocclusion.

Fig 10. Moments Mx, My and Mz on the canine, lateral incisor, and premolar with passive and elastic ligation, with 0.018-in NiTi wire in a high canine malocclusion.
In our pilot experiment, we attempted to explore the effect of the ligation method on the resultant orthodontic force system. Two identical malocclusions were simulated; in one, passive self-ligating brackets were used, and, in the other, conventional ligation was used. A high maxillary right canine was created on the OSIM, and 0.018-in NiTi wire was tied to all brackets. Three-dimensional force and moment data were collected as the canine was moved 4 mm occlusally and then back to the default position. Data on the high canine, lateral incisor, and first premolar (teeth mesial and distal of the canine) were obtained. In this high canine situation, the ideal force system is an extrusive force on the canine, with intrusive reciprocal forces on the premolar and lateral incisor and all other components kept at a minimum.

On examining the 3D force data, it is not surprising that the main difference between the 2 ligation methods is in the resistance to sliding (Fx forces) recorded on the lateral incisor and the premolar; this force was higher in the conventional ligation method compared with passive ligation.

However, this increased resistance to sliding considerably affected the other 5 components of the force system. The preliminary data here suggest that the 6 components of the force system are affected by the ligation method. Fz showed the most noticeable effect of the ligation method; passive self-ligation allowed the true load-deflection curve of the wire to be expressed. Flat loading and unloading curves are expected of a superelastic alloy, whereas the same wire produced a distinctly different load-deflection curve with conventional ligation (Fig 9). Adding elastomeric ties to the brackets increased the magnitude of the rest of the components (Fx, Fy, Mx, My, and Mz). Fx (mesiodistal forces) and Fy (buccolingual forces) are especially important, since the increase in Fx (resistance to sliding) seems to cause the increase in Fy (labiolingual forces). The total resistance to sliding in the brackets posterior to the maxillary right canine was less than the resistance mesial to the canine; this meant that the wire will preferentially slide through the canine, first premolar, and second premolar brackets rather than sliding through the brackets from the maxillary right lateral incisor to the left second premolar. This resulted in a net mesially directed force on the canine as it was moved into the high position; this force was reversed into a distal force halfway through the unloading phase for the same reason. The moments were also considerably affected by the ligation method (Fig 10). In a vertically displaced canine, the ideal force system would be an extrusive force on the canine with reciprocal unwanted forces and moments on the lateral incisor and first premolar. It is impossible to eliminate the unwanted components of this force system, but it seems that the unwanted forces and moments were considerably larger with conventional ligation compared with passive ligation.

We tried to simulate as closely as possible the oral environment, but elastomeric force decay is bound to have some effect on the force system.37,38 It would be an interesting test to induce force decay and age the elastomeric ties before using them and assess how that would affect the resultant force system. The elastomeric ties in this experiment were not changed during the simulated tooth movement. In future studies, we will replicate this test but change the elastomeric ties at every 1 mm of canine movement.

We are now modifying all the tooth connectors to install motorized micrometers on all teeth that can be controlled by the OSIM software; this will allow multiple simultaneous tooth movements with real-time data gathering. We will soon be able to have the OSIM model react to orthodontic forces and move the teeth accordingly while the data are recorded.

Our objective was not to accurately simulate the oral environment and control for all possible variables, including moisture, lip pressure, tongue pressure, PDL compliance, pressure distribution in the PDL, alveolar bone level and geometry, and many biologic variations. Our objective was to understand the biomechanical force system at the bracket-wire interface. PDL compliance causes minor movements in this interface and can play a role in determining the resultant force system. However, PDL simulation is a complicated process because of the wide variation of the data in the literature. In the future, we plan to use an elastomeric material or oil-immersed leather as an interface between the brackets and the load cells to simulate the viscoelastic properties of the PDL. The connectors on our device already have a small amount of compliance. We plan to measure the compliance of the new tooth connectors and determine how much more needs to be added to approximate true PDL compliance.

**CONCLUSIONS**

Orthodontic force systems resulting from contemporary orthodontic applications with full fixed appliances are considered indeterminate—ie, impossible to analyze and predict. Our aim was to build and validate a laboratory-based human mouth model capable of measuring forces and moments acting on teeth with multi-banded orthodontic fixed appliances. We were successful in building this model, which will allow us for the first time to determine with high accuracy the forces acting on orthodontically treated teeth.

Future research will focus on simulating many types of orthodontic clinical applications of full-fixed or
partial-fixed appliances. Gathering enough data (based on clinical records of treated patients) will pave the way toward building a solid computer model for orthodontic tooth movement simulation and prediction. Such a model will be based on actual experimental orthodontic biomechanical data.

When considering the overall force system generated in the high canine simulation, we can safely conclude that, based on these in-vitro measurements, the passive self-ligation method produced a more accurate force system for this malocclusion, with fewer unwanted forces and moments compared with elastic conventional ligation. Based on those findings, we might not be able to make definite predictions on the effect of these differences on the actual tooth movements. However, it is safe to conclude that different force systems produce different types of tooth movement; therefore, we would expect to see more vertical canine movement and less tipping of the adjacent teeth with passive ligation compared with conventional ligation.

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REFERENCES