Development and modification of a device for three-dimensional measurement of orthodontic force system: The V-bend system re-visited

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We developed a device to evaluate the orthodontic force applied by systems requiring high operability. A life-sized, two-tooth model was designed, and the measurements were performed using a custom-made jointed attachment, referred to as an “action stick”, to allow clearance for the oversized six-axis sensors. This tooth-sensor apparatus was accurately calibrated, and the error was limited. Vector analysis and rotating coordinate transformation were required to derive the force and moment at the tooth from the sensor readings. The device was then used to obtain measurements of the force and moment generated by the V-bend system. Our device was effective, providing results that were consistent with those of previous studies. This measurement device can be manufactured with force sensors of any size, and it can also be expanded to models with any number of teeth.

Keywords: Force sensor, Orthodontics, Force system, Moment, Measurement device

INTRODUCTION

The optimum magnitude of applied force is an important concern in orthodontics. In biological studies of the periodontal ligament (PDL) and alveolar bone, the magnitude of the applied force affected the type of bone remodeling: light forces led to “frontal resorption”, which generates smooth and continuous tooth movement, while heavy forces led to “undermining resorption”, which causes necrosis in a portion of the PDL and delays remodeling¹. In other words, heavier forces act to retard the velocity of tooth movement^{2,3}. Extremely heavy forces have been shown to increase the tendency of root resorption in many animal studies^{3–5}. Additionally, greater forces seemed to generate a greater pain reaction^{6,7}. Therefore, the optimal orthodontic force must be large enough to overcome the threshold of resistance from the PDL and bone and maximize the rate of tooth movement, but not too large to cause adverse effects.

The moment of the force acting on the tooth is another important issue, because the pattern of tooth movement is highly dependent on the moment-to-force ratio^{8–10}. The application of a single force on a crown will result in tooth tipping, so an additional counter moment is required to produce bodily movement or root movement. A small change in the moment-to-force ratio will result in a different clinical outcome¹¹. Although

moments can be generated easily using a multibracket appliance, accurate control remains a challenge.

A variety of force-generating systems are used in modern orthodontic treatment. In order to evaluate a system's performance, an objective standard is needed to determine whether the optimal forces and moments are being generated. This requires the measurement of both forces and moments in six spatial components. Multiple measurement devices have been developed for this purpose and can be roughly divided into two categories: those based on actual measurement, and those based on computer simulation. To obtain actual measurements, Friedrich *et al.* attempted to fabricate an apparatus that allowed intra-oral measurement^{12–14}; Lapatki *et al.* designed a smart bracket system with encapsulated microelectronic sensor chips^{15–17}; Badawi *et al.* developed an orthodontic simulator (OSIM) that incorporates multi-axis force sensors and an artificial dental arch¹⁸.

The OSIM has been used in several previous studies^{19–22} and has been shown to be highly effective for measuring the forces and moments experienced by the whole arch *in vitro*. However, these studies were performed only for a “straight” archwire with different artificial tooth positions and ligation methods. The original OSIM device consists of a large sensor apparatus mounted around the artificial dental arch, which

Fig. 1 A schematic diagram of the original orthodontic simulator (OSIM) design (a). Note that the artificial dental arch is placed inside the sensors and other auxiliary parts (*i.e.*, the tooth location monitors, sensor cables, and amplifier). The operability is decreased because these parts are obstructive when delicate orthodontic procedures are required. Our modified device is shown as a schematic diagram (b) and a photograph (c). Tooth A was assumed to be a distal tooth, and tooth B was a mesial tooth. The X, Y, and Z axes were defined along the mesio-distal direction (mesial: +, distal: –), the bucco-lingual direction (lingual: +, buccal: –), and the occluso-gingival direction (occlusal: +, gingival: –), respectively. The directions of the moments are defined by the right-hand rule.

limits the accessibility of the arch when orthodontic adjustments are required (Fig. 1a). Therefore, the analysis of orthodontic force systems with multiple wire-bending patterns or loop designs that require frequent wire application and removal is difficult with the OSIM. The aim of this study was to modify the OSIM design and assess its effectiveness and reliability in acquiring measurements from a V-bend system.

MATERIALS AND METHODS

Design of measuring device

The device consists of two six-axis sensors, 18 mm in diameter (CFS018CA101AS, Leptrino, Nagano, Japan), and two cylindrical artificial teeth composed of an acrylic compound (Veroclear rgd810, Stratasys, MN, USA) and fabricated by a 3D printer. The artificial teeth were incorporated to serve as a model of two life-sized adjacent lower premolars with which to test our device. The sensors and teeth were connected by a custom-made joint that we refer to as an “action stick” (Fig. 1b and c). The action stick was necessary because the sensors are too large to fit within the dental arch together (Fig. 2). The action stick was designed to be positioned oblique

to the tooth axis, thus providing sufficient clearance for the sensors. They were composed of aluminum to ensure sufficient rigidity. The dimensions of the action stick are described in detail in Fig. 3. The action stick and force sensor were linked to an aluminum slider and plate. These sliders allowed the position of the tooth-sensor apparatus to be adjusted as necessary.

Self-ligated brackets (mini Clippy; 0.018×0.025-inch slot for the upper premolar, angulation 0°, torque –7°; TOMY, Tokyo, Japan) were bonded to the teeth using a self-curing dental adhesive resin cement (Orthomite Super-Bond, Sun Medical, Shiga, Japan). We used self-ligating brackets to avoid biases during ligation; these orthodontic materials are commonly used in clinics in Japan. The two bracket slots were oriented in a straight line, parallel in all three planes, using a straight full-thickness stainless steel wire as a guide during the bonding procedure. The inter-bracket distance (designated as “L” here, measured from the center of the brackets) was set at 8 mm, according to the typical mesio-distal dimension of the lower premolars in the Japanese population²³. Table 1 defines the six axes that were used in our experimental model of the right lower premolars.

Fig. 2 The dental arch placed in the inner circumference (a) and on the outer circumference (b) of the sensor apparatus. Since the sensor was larger than the artificial teeth (18-mm sensor diameter versus 6-mm teeth in this study), the sensors occupy too much space inside the dental arch, thus causing spatial conflict as the sensors impinge on each other when applied directly to the arch.

Fig. 3 The height (26.8 mm) and length (50 mm) of the action stick are shown. Note that the tooth axis was rotated 15° clockwise for tooth A and 15° counter-clockwise for tooth B relative to the coordinates of the sensors.

Table 1 Coordinate definitions corresponding to the orthodontic terms

Axis	Plus	Minus	Orthodontic direction
Fx	Mesial force	Distal force	—
Fy	Lingual force	Buccal force	First order
Fz	Occlusal force	Gingival force	Second order
Mx	Crown-buccal torque	Crown-lingual torque	Third order
My	Mesial tipping moment	Distal tipping moment	Second order
Mz	Mesial rotation moment	Distal rotation moment	First order

Calibration procedure

The sources of error in this device could be classified as the errors associated with the sensor itself, and the errors associated with the transmission of the force and moment through the action stick. The former is shown

in Table 2, which was provided by the manufacturer; the latter was determined by a calibration procedure of our devising. This procedure was performed under a test stand (FGS-5TV, NIDEC-SHIMPO, Kyoto, Japan) that can generate forces through an exclusive jig and

Table 2 Specifications of the six-axis force sensor, CFS018CA201a

	Fz	FyFy	MxMyMz
Rated capacity (R.C.)	±50 N	±50 N	±100 N·m
Allowable load		±150% R.C.	
Non linearity		±2.0% R.C.	
Interference		±3.0% R.C.	
Resolution		±1/2,000	

“Non-linearity” describes the error between loading and output; “Interference” describes the output of the axes without loading. These terms comprise the index of precision.

Fig. 4 A photograph of the test stand. The force gauge was fixed on a slide and was able to deliver an arbitrary magnitude of force to the tooth-sensor apparatus.

force gauge (FGP-0.5, NIDEC-SHIMPO) (Fig. 4). The tooth-sensor apparatus was fixed on the test stand, and the jig automatically applied a force on the sensor. The specifications of the force gauge are shown in Table 3. We applied loads from 0 to ±5 N in increments of 0.5 N and collected data along three axes. The calibration was performed at room temperature (27.8°C), which was within a permissible range for both the six-axis sensor and the force gauge. These data were then used to calculate the six components of the error, which are

Table 3 Specifications of the force gauge

	FGP-0.5
Rated capacity (R.C.)	±5.000N
Precision	±0.2% R.C.
Limited overload	200% R.C.
Resolution	0.001 N

reported in Tables 4 and 5. These calculations were based on the method and theory described in the original OSIM study¹⁸⁾. The overall errors were 2.06% for force and 2.0% for moment. These values indicate that the error was limited and was equivalent to the error rates reported for the original OSIM.

Vector analysis

Although we are interested in measuring the forces and moments acting on the tooth, the sensor readings cannot be used directly due to the presence of the action stick. A transformation function must be applied to derive the actual moments generated at the tooth from the measurements that are detected by the sensor (defined as the center of the bracket). Figure 5 shows two arbitrary points, P and Q , for an immobile rigid body, with a distance r between them. Assuming that the body is in equilibrium,

$$\vec{F}_P = \vec{F}_Q.$$

This means that the force vectors need no transformation. However, for point Q , the moment will become

$$\vec{M}_P = \vec{M}_Q - (\vec{F}_P \times \vec{r})$$

because $\vec{F}_P$ will generate a moment $\vec{F}_P \times \vec{r}$ on Q .

Applying this to our device, the transformation formulas become:

$$\vec{F}_{tooth} = \vec{F}_{sensor}$$

and

Table 4 Force errors of the tooth-sensor apparatus

Force sensor	Fx Load applied			Fy Load applied			Fz Load applied			Average error in	Average error in	Average error in	Overall force error per
	Fx	Fy	Fz	Fx	Fy	Fz	Fx	Fy	Fz	Fx per	Fy per	Fz per	
Average error (%)	1.36	2.80	1.84	1.99	2.16	2.59	1.50	2.75	1.53	2.36	1.64	3.08	—
Per Axis SD (%)	0.96	0.63	0.83	1.04	1.11	1.07	0.53	1.00	0.94	0.84	1.37	1.22	—
Average force error per loaded axis (%)	—	2.00	—	—	2.25	—	—	1.93	—	—	—	—	2.06

Table 5 Moment errors of the tooth-sensor apparatus

Force sensor	Fx Load applied			Fy Load applied			Fz Load applied			Average error in	Average error in	Average error in	Overall moment error per
	Mx	My	Mz	Mx	My	Mz	Mx	My	Mz	Mx per	My per	Mz per	
Average error (%)	1.11	1.49	1.55	5.38	0.87	0.46	2.56	3.11	1.47	2.17	1.11	0.03	—
Per Axis SD (%)	1.07	0.95	0.86	1.82	0.70	0.73	1.70	0.71	1.03	1.53	0.79	0.87	—
Average moment error per loaded axis (%)	—	1.38	—	—	2.24	—	—	2.38	—	—	—	—	2.00

Fig. 5 A representative rigid body with two arbitrary points, P and Q . The moment and force at P and Q are shown.

$$\overrightarrow{M_{tooth}} = \overrightarrow{M_{sensor}} - (\overrightarrow{F_{tooth}} \times \vec{r})$$

where F_{tooth} and M_{tooth} are the force acting on, and the moment about the center of the bracket, respectively, F_{sensor} and M_{sensor} are the force acting on, and the moment about the center of the sensor, respectively, and r is the distance between the two points.

Coordinate transformation

Along with the vector transformations, a transformation of the coordinate system was also required, as the coordinate system for the measured data was not coincident with the tooth axes. The principle of this

transformation is described in the equations below. Starting from a three-dimensional coordinate system with axes X , Y , and Z , we rotated these axes through angles θ_x , θ_y , and θ_z , respectively, thus producing a new set of axes, X' , Y' and Z' (Fig. 6a). To determine the new coordinates, we first consider the transformation in two dimensions:

$$\begin{aligned} x' &= x \cos \theta - y \sin \theta \\ y' &= x \sin \theta + y \cos \theta \end{aligned}$$

$$\begin{pmatrix} x' \\ y' \\ 1 \end{pmatrix} = \begin{pmatrix} \cos \theta & -\sin \theta & 0 \\ \sin \theta & \cos \theta & 0 \\ 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x \\ y \\ 1 \end{pmatrix}, \text{ where the lowercase}$$

letters represent the coordinate (Fig. 6b). Then, we expand this formula to three dimensions (Fig. 6c). For the YZ plane,

$$R_1: \begin{pmatrix} x' \\ y' \\ z' \\ 1 \end{pmatrix} = \begin{pmatrix} 0 & 0 & 0 & 0 \\ 0 & \cos \theta_y & -\sin \theta_y & 0 \\ 0 & \sin \theta_y & \cos \theta_y & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x \\ y \\ z \\ 1 \end{pmatrix}$$

for the XZ plane,

$$R_2: \begin{pmatrix} x' \\ y' \\ z' \\ 1 \end{pmatrix} = \begin{pmatrix} \cos \theta_x & 0 & \sin \theta_x & 0 \\ 0 & 0 & 0 & 0 \\ -\sin \theta_x & 0 & \cos \theta_x & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x \\ y \\ z \\ 1 \end{pmatrix}$$

and for the XY plane,

Fig. 6 A perpendicular coordinate system (X, Y, and Z) was rotated to produce new coordinates (X', Y', and Z') in three dimensions (a). A simpler situation for two-dimensional rotation (b). Decomposition of three-dimensional rotation into three planes (c).

$$R_3: \begin{pmatrix} x' \\ y' \\ z' \\ 1 \end{pmatrix} = \begin{pmatrix} \cos \theta_x & -\sin \theta_x & 0 & 0 \\ \sin \theta_x & \cos \theta_x & 0 & 0 \\ 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x \\ y \\ z \\ 1 \end{pmatrix}$$

resulting in a rotation matrix. Thus, the overall transformation is given by

$$\begin{pmatrix} x' \\ y' \\ z' \\ 1 \end{pmatrix} = R_1 R_2 R_3 \begin{pmatrix} x \\ y \\ z \\ 1 \end{pmatrix} = \begin{pmatrix} 0 & 0 & 0 & 0 \\ 0 & \cos \theta_y & -\sin \theta_y & 0 \\ 0 & \sin \theta_y & \cos \theta_y & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} \cos \theta_x & -\sin \theta_x & 0 & 0 \\ \sin \theta_x & \cos \theta_x & 0 & 0 \\ 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x \\ y \\ z \\ 1 \end{pmatrix}$$

With this relationship, the corresponding position of any given coordinate in the new coordinate system can be derived, as long as the original position and the angles between the axes are known.

Calculation procedure

To derive the final values from the sensor readings, the calculations involved a combination of vector analysis and rotational coordinate transformation. To demonstrate, we provide a concrete example. Supposing that the measured values of the sensor are

$$F_x = F_z = 0.2 \text{ N}, M_x = 30 \text{ N} \cdot \text{mm}.$$

The dimensions of the action stick are as shown in Fig. 3, thus

$$r = \begin{pmatrix} r_x \\ r_y \\ r_z \end{pmatrix} = \begin{pmatrix} 0 \\ 50 \\ 26.8 \end{pmatrix}.$$

According to the vector analysis, we know that

$$\begin{pmatrix} F_{x, \text{tooth}} \\ F_{y, \text{tooth}} \\ F_{z, \text{tooth}} \end{pmatrix} = \begin{pmatrix} F_{x, \text{sensor}} \\ F_{y, \text{sensor}} \\ F_{z, \text{sensor}} \end{pmatrix} = \begin{pmatrix} 0.2 \\ 0 \\ 0.2 \end{pmatrix}$$

$$\begin{pmatrix} M_{x, \text{tooth}} \\ M_{y, \text{tooth}} \\ M_{z, \text{tooth}} \end{pmatrix} = \overrightarrow{M_{\text{sensor}}} - (\overrightarrow{F_{\text{tooth}}} \times \vec{r}) = \begin{pmatrix} 30 \\ 0 \\ 0 \end{pmatrix} - \begin{vmatrix} \vec{i} & \vec{j} & \vec{k} \\ 0 & 50 & 26.8 \\ 0.2 & 0 & 0.2 \end{vmatrix} = \begin{pmatrix} 20 \\ -5.36 \\ 10 \end{pmatrix}$$

where $\vec{i}$, $\vec{j}$, and $\vec{k}$ are unit vectors of the X, Y, and Z axes, respectively.

Then, we need to consider the rotation transformation. Because the rotation of the device was only 15 degrees in the XY plane,

$$\theta_x = \theta_y = 0, \theta_z = 15, \text{ so}$$

$$R_1 = \begin{pmatrix} 0 & 0 & 0 & 0 \\ 0 & \cos \theta_y & -\sin \theta_y & 0 \\ 0 & \sin \theta_y & \cos \theta_y & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} = \begin{pmatrix} 0 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix}$$

$$R_2 = \begin{pmatrix} \cos \theta_x & 0 & \sin \theta_x & 0 \\ 0 & 0 & 0 & 0 \\ -\sin \theta_x & 0 & \cos \theta_x & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} = \begin{pmatrix} 1 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix}$$

$$R_3 = \begin{pmatrix} \cos 15 & -\sin 15 & 0 & 0 \\ \sin 15 & \cos 15 & 0 & 0 \\ 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} = \begin{pmatrix} 0.966 & -0.259 & 0 & 0 \\ 0.259 & 0.966 & 0 & 0 \\ 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix}.$$

Thus, the force on the tooth will be

$$\begin{pmatrix} F_{x, \text{tooth}} \\ F_{y, \text{tooth}} \\ F_{z, \text{tooth}} \\ 1 \end{pmatrix} = R_1 R_2 R_3 \begin{pmatrix} F_{x, \text{tooth}} \\ F_{y, \text{tooth}} \\ F_{z, \text{tooth}} \\ 1 \end{pmatrix} = \begin{pmatrix} 0.19 \\ 0.05 \\ 0.2 \\ 1 \end{pmatrix},$$

and the moment of the tooth will be

$$\begin{pmatrix} M_{x, \text{tooth}} \\ M_{y, \text{tooth}} \\ M_{z, \text{tooth}} \\ 1 \end{pmatrix} = R_1 R_2 R_3 \begin{pmatrix} M_{x, \text{tooth}} \\ M_{y, \text{tooth}} \\ M_{z, \text{tooth}} \\ 1 \end{pmatrix} = \begin{pmatrix} 20.7 \\ 0 \\ 10 \\ 1 \end{pmatrix}$$

These analysis procedures were incorporated into a

program so that we could derive the moment and force in real-time.

Experimental procedure

The V-bend system was evaluated in this study. Two common types of commercial orthodontic wires, stainless steel (SS) (Unitek™ Stainless Steel Straight Lengths; 0.016×0.022 inches; 3M Unitek, Monrovia, CA, USA) and beta titanium (TMA) (TMA LOW FRICT; 0.016×0.022 inches; ORMCO, Orange, CA, USA), were tested. For each material, three samples were measured. In addition, only the straight portion at the end of the preformed wire was used for the TMA wires. The degree of the V-bend was 20° in all cases. These wire samples were then set into bracket slots by passive ligation (Fig. 7)

Hereafter, the distal artificial tooth is referred to as tooth A, and the mesial artificial tooth is referred to as tooth B. Measurement was performed when the V-bend was located at 1/4L, 1/3L, 1/2L, 2/3L, and 3/4L of the inter-bracket distance between the two teeth. (2, 2.67, 4, 5.33, and 6 mm from the center of the bracket on tooth A). Before every change in the bend position, the wire sample was removed from the bracket and measured to

Fig. 7 Experimental setup with the V-bend. A bent wire was set in the center of tooth A and tooth B.

confirm that the degree of the V-bend stayed constant. All procedures were performed by the same operator, who was a trained orthodontist.

RESULTS

Three wire samples and 5 bend positions, for a total of 15 data sets, were evaluated for each material (SS and TMA). The average and standard deviation of the three samples were calculated. The V-bend tested in this study was activated to give a second-order (mesial or distal tipping) moment (M_y); vertical forces (F_z) were also generated due to equilibrium, and the tendencies of these two main components are shown in a graph (Fig. 8).

The measurements varied according to the bend location: at 1/2L, F_z was 0 for both teeth and the absolute values of M_y were identical, though the directions were opposite; at 1/3L and 2/3L, the value of M_y was close to 0 for the nearer tooth; at 1/4L and 3/4L, the moment was slightly reversed from the original direction of bend activation. These results were consistent with the findings in previous reports²³⁻²⁶. The forces and moments generated by the TMA wire were less than half the values obtained with the SS wire, because the stiffness of TMA is approximately 40% that of SS²⁷.

The values for all six components are reported in Tables 6–9. The activation was designed for application only in the second-order direction, thus F_x and F_y showed a limited magnitude. However, M_x and M_z showed large values when M_y increased. The increased moment is most likely the result of the loss of wire-bracket play. The larger standard deviation was considered to be the result of subtle differences in the wire setting.

DISCUSSION

The incorporation of the force sensor is a key consideration in the development of a device for the measurement of orthodontic forces and moments. Rosarius and Friedrich's system¹²⁻¹⁴ linked sensors to the multibracket side of the device instead of to the tooth side to allow *in vivo* measurement. In their design, the use of a divisible joint between the brackets and teeth

Fig. 8 The results of the V-bend tests. Only the second-order moment (M_y) and vertical force (F_z) are shown. SS: stainless steel; TMA: beta titanium.

Table 6 Forces and moments on tooth A with TMA wires

		Fx (N)		Fy (N)		Fz (N)		Mx (N•mm)		My (N•mm)		Mz (N•mm)	
		Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD
V-bend position	1/4L	0.16	0.04	-0.28	0.17	2.10	0.03	-16.52	0.90	-16.47	0.56	-13.23	1.40
	1/3L	0.18	0.01	-0.05	0.02	1.48	0.13	-10.81	0.21	-11.83	0.99	-3.76	1.55
	1/2L	0.00	0.03	-0.14	0.17	0.06	0.09	-3.23	1.38	-4.88	0.13	1.25	1.20
	2/3L	-0.01	0.05	-0.12	0.16	-0.88	0.22	0.14	0.66	0.41	1.00	3.20	1.32
	3/4L	0.10	0.11	-0.12	0.20	-1.68	0.05	2.45	1.05	0.73	0.24	8.41	0.42

Table 7 Forces and moments on tooth B with TMA wires

		Fx (N)		Fy(N)		Fz (N)		Mx (N•mm)		My (N•mm)		Mz (N•mm)	
		Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD
V-bend position	1/4L	-0.17	0.10	0.31	0.12	-2.21	0.03	8.71	0.69	-0.75	0.12	-5.34	0.43
	1/3L	-0.07	0.02	0.03	0.02	-1.53	0.12	4.03	0.90	1.33	0.14	-4.22	0.52
	1/2L	-0.12	0.18	0.05	0.05	-0.05	0.10	-2.51	0.37	4.96	0.55	-0.51	1.14
	2/3L	-0.16	0.11	-0.07	0.13	0.91	0.23	-7.24	0.91	8.08	0.40	3.59	0.73
	3/4L	-0.27	0.13	-0.26	0.19	1.73	0.04	-12.84	0.79	14.08	0.33	5.72	0.87

Table 8 Forces and moments on tooth A with SS wires

		Fx (N)		Fy (N)		Fz (N)		Mx (N•mm)		My (N•mm)		Mz (N•mm)	
		Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD
V-bend position	1/4L	0.48	0.23	-0.32	0.06	4.82	0.12	-38.98	1.31	-39.17	2.11	-30.52	0.18
	1/3L	0.01	0.14	-0.25	0.06	3.93	0.26	-33.02	2.97	-34.24	3.14	-24.87	3.33
	1/2L	0.14	0.33	-0.03	0.21	0.14	0.17	-11.88	0.09	-15.72	1.22	-1.42	0.67
	2/3L	-0.30	0.32	0.10	0.14	-3.28	0.33	3.28	4.26	0.61	3.73	22.01	4.42
	3/4L	-0.24	0.14	0.12	0.03	-3.85	0.35	9.39	2.06	5.37	2.54	27.74	2.96

Table 9 Forces and moments on tooth B with SS wires

		Fx (N)		Fy (N)		Fz (N)		Mx (N•mm)		My (N•mm)		Mz (N•mm)	
		Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD	Ave	SD
V-bend position	1/4L	-0.13	0.05	0.46	0.20	-5.11	0.11	17.30	0.56	-0.73	0.21	-8.97	1.79
	1/3L	0.06	0.07	0.67	0.04	-4.17	0.28	10.68	1.42	2.85	0.26	-6.54	0.41
	1/2L	-0.07	0.32	-0.09	0.17	-0.18	0.15	-7.01	1.33	14.91	0.48	-0.02	1.86
	2/3L	-0.08	0.25	-0.40	0.29	3.35	0.37	-31.73	2.12	31.08	0.37	11.68	0.81
	3/4L	-0.20	0.07	-0.58	0.07	3.97	0.35	-34.11	1.86	33.36	1.26	14.77	0.54

was necessary because the force and moment could not be transmitted to the sensors under a mechanical equilibrium. The concept of this device was interesting,

though the system was ethically questionable.

Three-axis sensors²⁸⁾ or stress sensor chips can be fabricated to be thin and are as small as the real tooth.

These sensor chips can be applied under the base surface of brackets. However, moment sensing is not possible with the three-axis force sensor. The smart bracket system^{15–17}, incorporating multiple micro-sensor chips, is an excellent innovation that successfully transforms the stress signal to the moment component utilizing the unique “stress-fingerprint” technique, though it is unlikely that this approach will become widely adopted in orthodontic mechanics research due to its high cost. Six-axis sensors (or force transducer and load cell) allow for moment measurement. Though the size of these sensors is being reduced, the currently available ones are still larger than real teeth. Connecting the sensors directly to artificial teeth and brackets using a straight stick is an intuitive and common method^{29,30}. Nevertheless, these systems cannot be expanded to the full arch because multiple sensors will impinge on each other. The OSIM was thus developed to enable the simultaneous measurement of all teeth through a scaffold that secures multiple sensors on the external surface of an artificial dental arch. The OSIM was developed with a focus on the adjustment of the location of the artificial teeth; in other words, with the goal of studying malocclusion of the teeth in version or the arch form of preformed archwires. However, this approach reduces operability in the dental arch, as described in the introduction.

On the other hand, although the straight-wire technique and the use of a pre-adjusted bracket system is becoming popular in contemporary orthodontics, plain wire is not appropriate in all clinical situations, and the mechanics described here are effective for local, fine control of tooth movement. These mechanics, such as the use of loop mechanics for bodily movement on extraction space closure, a cantilever spring for uprighting a tipped molar, a utility arch for incisor intrusion¹, *etc.*, should be tested and measured to confirm their effectiveness, which entails the development of a valid method of measurement. Moreover, the use of a life-sized model is

effective for training junior orthodontists to reproduce standard bending.

Therefore, in this study, our main purpose was to modify the OSIM such that it could be conveniently used for orthodontic devices that incorporate the bended wire or loop, which requires setting the wire into the slot and removing it multiple times. The main feature of our design was the use of a modified L-shaped linking structure (referred to as the “action stick” here) that allowed us to place the sensors inside the artificial arch, leaving the external surface exposed, thus increasing the operability. The introduction of the action stick required several matrix transformations to derive the actual physical values experienced by the tooth from the sensor readings; these equations have not been described in previous studies. Though a model with only two teeth was used in the present study, our approach will be helpful in future investigations. Because the angle, length, and height of the action stick can be adjusted, our design can be applied to models with any number of artificial teeth (Fig. 9). However, when using our device, any alterations in tooth position would require recalculation of the new coordinate system, which can be time-consuming. Therefore, investigators should consider the objectives of their study when deciding which OSIM design is more appropriate, in other words, whether the device needs to accommodate multiple wire adjustments or repeated variations in the tooth position.

Taking measurements through action sticks introduces another source of error, and the precision of our measuring device inevitably declined slightly, compared to the original. Nevertheless, as the stiffness of the action sticks increased, the error in the measurement of overall force and moment was reduced and was considered to be acceptable for orthodontic mechanical analysis. In fact, when we used this device to evaluate the V-bend system, the measured values were equivalent to the theoretical values. Moreover, our

Fig. 9 A theoretical expanded design for our device, designed to measure half of the dental arch (a). Sensors are placed in a fan shape (b) because the heights of the action sticks vary slightly (c).

Note that within this design, the vector analysis was the same for all teeth because the distances from the sensor to the tooth were the same. However, the rotated coordinate transformation must be calculated independently because the rotated angle differs. The error of each tooth-sensor apparatus must be measured individually during the calibration procedure to ensure the accuracy of the whole device.

device was still effective, even in the comparatively low-stiffness material that produced lower magnitude forces. Since a force magnitude less than 50 g (0.5 N) is not relevant in most clinical situations, the ability to detect such small forces is not useful. In addition, because an increase in the moment (or torque) expression was observed in this study for all six components measured, it is possible to investigate the force system in the other direction, which was not initially predicted.

CONCLUSIONS

- A modified design of an orthodontic measuring device allowing greater operability on wire settings was evaluated.
- Through the introduction of the action stick, the problem of the sensor size was resolved.
- The transformation equation and theory for this design was detailed.
- The precision of this device was acceptable and the effectiveness was confirmed by testing the V-bend.
- This measuring device could be expanded to evaluate different numbers of teeth.

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