Jaw Bite Force Measurement Device

Dennis Flanagan, DDS[1]*
Horea Ilies, PhD[2]
Brendan O'Brien, BS[2]
Anne McManus, BS[2]
Beau Larrow, DDS[2]

We describe a cost-effective device that uses an off-the-shelf force transducer to measure patient bite force as a diagnostic aid in determining dental implant size, number of implants, and prosthetic design for restoring partial edentulism. The main advantages of the device are its accuracy, simplicity, modularity, ease of manufacturing, and low cost.

Key Words: dental implant, jaw force, bite force, edentulism, dental prosthesis

INTRODUCTION

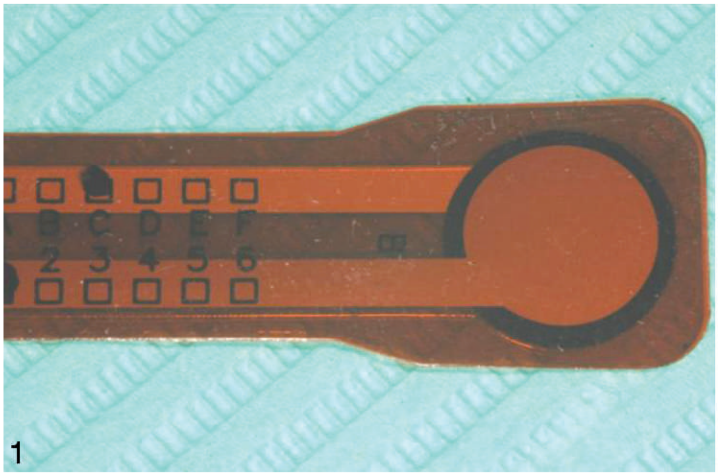
Dental implants are placed to enable restoration of patients' physiologic function. Implants support and/or retain prostheses, and the occlusal forces are transmitted to the implants and subsequently to the supporting bone. Clinical dentists rely on clinical judgment to determine implant size and prosthetic design. However, this is a qualitative judgment based on experience and education. At the same time, the range of biting forces of presenting patients is large, typically ranging from 500 to 1500 N.[1] Clearly, the selection of implant size and prosthetic design is influenced by the patient's jaw-force capability; thus, knowing the patient's jaw-force capability a priori can help the implant dentist ensure a favorable treatment outcome. This article describes a device designed to measure patient jaw force that is accurate, simple, and highly modular. Furthermore, the simplicity of its design favors efficient manufacturing processes and, therefore, suggests that the final product will be cost-effective for intraoral use.

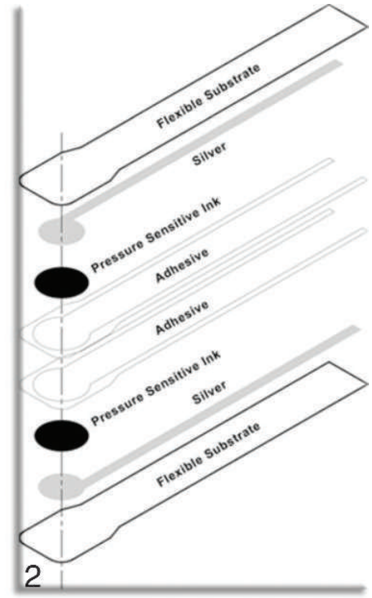
[1] Private Practice, Willimantic, Conn.
[2] Department of Mechanical Engineering, University of Connecticut, Storrs, Conn.
* *Corresponding author,* e-mail: dffdds@comcast.net
DOI: 10.1563/AAID-JOI-D-10-00101

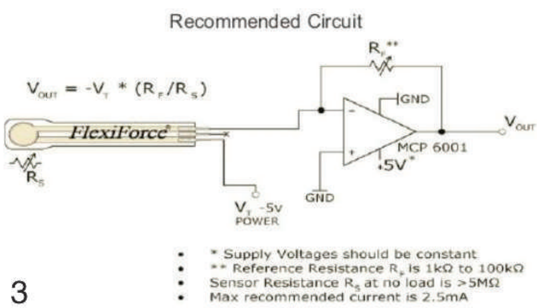
MATERIALS AND METHODS

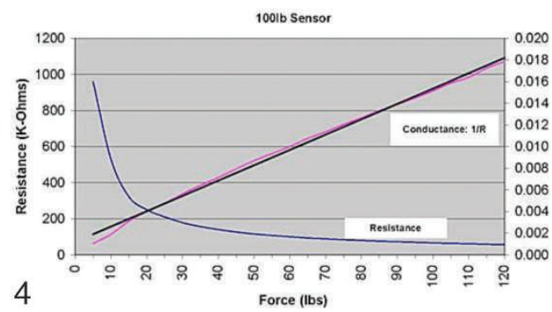
The device reported here represents the physiological use of an industrial mechanical measuring device that was enhanced for human oral usage. The Flexforce force transducer (Tekscan, South Boston, Mass) is a very thin flexible sensor that can measure up to 4000 N (Figure 1). It is used to measure forces between 2 solid or semisolid pieces. The piezoresistive sensor consists of layers of materials that produce a measurable change in electronic resistance under load (Figure 2).

The outer layer is a flexible envelope that seals the contents of the device from exogenous liquids. The seal, however, is not created for internal human use or multiple or long-term underwater uses. The next layer is a layer of silver on top of a layer of pressure-sensitive ink. The silver-ink acts as a resistor in a circuit (Figure 3). Initially, the resistance in the sensor is very high but when a force is applied the resistance changes. The resistance decreases with an increasing load, enabling a measurement of up to 4400 N. The change







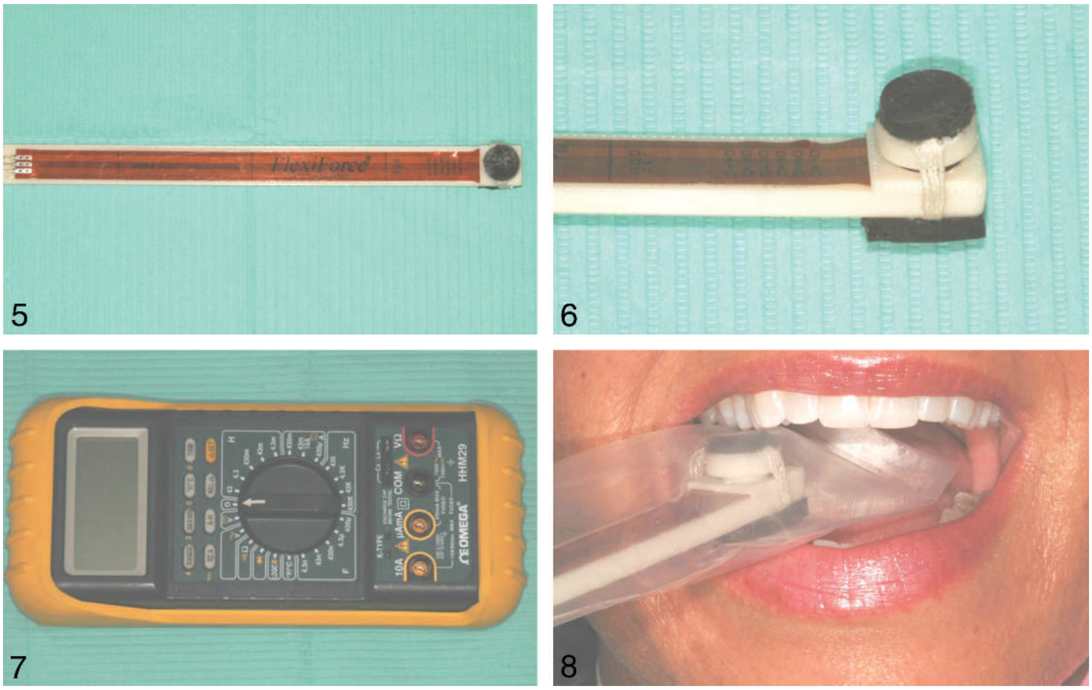


Figures 1–4. Figure 1. The sensor head of the industrial pressure-measuring device (Flexforce). **Figure 2.** The sensor is constructed in layers consisting of a flexible substrate layer; a silver layer; a pressure-sensitive ink layer; 2 layers of adhesive; and lower layers of ink, silver, and flexible substrate. **Figure 3.** Schematic of the electronic circuit of the sensor. **Figure 4.** An increase in applied bite force causes a decrease in electronic resistance, which is correlated with bite force in Newtons. There is a linear increase in conductance and an increase in applied bite force.

in resistance (R) is not linear, but the electronic conductance (1/R), however, is linear (Figure 4).

Applied forces below the limit do not damage the device, so it can be used repeatedly up to a million cycles. The sensor can withstand −9 to 60°C, and a high-temperature device is available with range of −9 to 204°C. The current cost is about $41. The error range of this device is reported to be 1.2–3.6%. A less expensive ($17) type of sensor is available, but the error range is 3–5%, although that may still be acceptable. This error range may not be clinically significant. The industrial sensor was incorporated into a device for intraoral human use.

A neoprene disc is attached to a laboratory-fabricated acrylic plastic button with cyanoacrylic adhesive and secured to fit onto the sensor to provide a platform for the opposing teeth to bite into (Figures 5 and 6). Neoprene plastic was chosen because it is soft and does not permanently deform after a bite episode. In addition, when tested by some of the authors (B.O., A.M., B.L.), it felt more comfortable than cloth, nitrile, or silicone. The disc-button sensor assembly is held together with laced Dacron tape (Fyr-Lace, 3M, St Paul, Minn). The height of the assembly is 18 mm, which seems to be the most comfortable size for the application of full jaw force. The fiber disc-button provides a substrate for jaw leverage and a relatively soft surface to prevent dental



Figures 5–8. Figure 5. The sensor was equipped with a laboratory-fabricated button and a neoprene disc. **Figure 6.** The button and discs were held with a Dacron lace. **Figure 7.** The sensor was connected to a multimeter (Omega Engineering) for readings, which were converted to bite force in Newtons. **Figure 8.** The assembled device is placed between the teeth, and the patient is asked to bite with maximum force.

injuries as it transfers the force to the sensor.

Because teeth can be cusped and pointed, and because of the inherent variation in locating the sensor in a patient's mouth, highly repeatable measurements of patient bite force are improbable. However, the repeatability of the device in lab measurements is smaller than 3.5%. The button provides a smooth uniform surface against the sensor during the force application to ensure accuracy.

A disposable 2″ × 12″ 4 mil polyethylene plastic bag sleeve (Universal Pacific, Ontario, Calif) covers the device to allow for convenient reusability. This device may be autoclavable, but this was not tested. Autoclaving pressure and humidity may cause leaks in the adhesive sealer of the sensor. The sensor is connected to a multimeter (Omega Engineering, Stamford, Conn) to detect the resistance changes, and this was converted to Newtons of force using a computational table of values (Figure 7) (Table). Obviously, other principles could be used for reading the force directly on the device. For example, the device could be augmented by a digital display and a programmable chip that would convert volts to Newtons. However, for the purposes of this study, which was focused on designing a simple, accurate, and cost-effective device, our force reading mechanisms provided a simple, robust, and quick solution. The device is placed between the teeth in the area to be recorded and the patient is asked to bite into the device at full jaw capacity (Figure 8). The patient is asked to apply the most jaw force he or she is able to.

Discussion

Dental implants support and retain oral prostheses and thus help restore function for edentulous patients. Occlusal jaw forces are transmitted through the prosthesis to the bone. The bone interface of the osseointegrated implant may tolerate

TABLE

The table presents the mathematical conversion of the measured resistance changes into Newtons of force generated by the patient

Resistance in Kilo Ohms	Force (N)
14.5	2023
15.0	1908
16.0	1707
17.0	1539
18.0	1398
20.0	1173
22.0	1005
24.0	875
26.0	772
28.0	690
30.0	622
32.0	566
34.0	519

up to 150 N of lateral force before a micromovement is induced and integration is lost.[2] The magnitude of the generated jaw force may be sufficient to cause a micromovement in some patients.

Each patient has an individual maximum bite force that may influence the location, size, and number of implants and the prosthetic design. Although most clinicians do not use quantitative values to design implant supported restorations, knowledge of a maximum jaw-force measurement may allow the clinician to appropriately design the treatment to prevent failure after loading a definitive prosthesis. Patients do not consistently and reliably produce the same maximum bite force.[3] Thus, a truly definitively accurate measurement may be academic.

This particularly designed device may not be easily used for completely edentulous patients. Measuring a the jaw force of a patient receiving a dental implant may allow for a clinical calculation to predict the feasibility of immediate loading. The patient's maximum generated jaw force would not be enough to cause a micromovement of the newly placed implant. Micromovement may cause enough of a luxation of the implant for a microhemorrhage and subsequent fibrous tissue formation and loss of or cessation of osseointegration. This would be based on and related to the selected implant's size, location, length, and number as well as the measured jaw force. Consequently, if a patient were able to generate a large magnitude bite force, then that patient may not be a candidate for an immediately truly functionally loaded prosthesis. However, a patient who generates a very low magnitude of occlusal force may be appropriate for a truly functionally loaded prosthesis. This would depend on the location in the arch for mechanical advantage, the number and size of the implants placed, and the bone density and cortical thickness of the site. This device could be used to perform measurements that could answer some of these questions.

CONCLUSIONS

A cost-effective device was assembled to measure patient maximum bite force as a diagnostic aid for determining dental implant size, number of implants to be used for prosthetic support, and design.

REFERENCES

1. Brunski JB. In vivo bone response to biomechanical loading at the bone/dental implant interface. *Adv Dent Res*. 1999;13:99–119.

2. Flanagan DF, Ilies H, Lasko B, Stack J. Force and movement of non-integrated implants: an in vitro study. 2008;34:128–134.

3. Roldan S, Buschang PH, Isaza Salarriaga JF, Throckmorton G. Reliability of maximum bite force measurements in age-varying populations. *J Oral Rehabil*. 2009;36:801–807.