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TECHNICAL NOTE

A CONDUCTIVE POLYMER SENSOR FOR MEASURING EXTERNAL FINGER FORCES

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Abstract—This paper describes the construction and use of a durable and thin force sensor that can be attached to the palmar surface of the fingers and hands for studying the biomechanics of grasp and for use in hand injury rehabilitation. These force sensors were constructed using a modified commercially available conductive polymer pressure sensing element and installing an epoxy dome for directing applied forces through a 12 mm diameter active sensing area. The installation of an epoxy dome was effective for making the sensors insensitive to contact surfaces varying from 25 to 1100 mm^2 and a 16 mm radius surface curved convex towards the finger. The completed sensors were only 1.8 mm thick and capable of being taped to the distal phalangeal finger pads. They were calibrated on the hand by pinching a strain gage dynamometer. The useful range was between 0 and 30 N with an accuracy of 1 N for both static loading and normal dynamic grasp activities. The sensor time constant was 0.54 ms for a step force input. Because of varying offset voltages every time the sensors were attached, these sensors should be calibrated on the hand before each use. The sensors were used for measuring finger forces during controlled pinching and lifting tasks, and during ordinary grasping activities, such as picking up a book or a box, where the useful force range and response for these sensors were adequate.

INTRODUCTION

Practical finger and hand force sensors are needed for hand biomechanics research, clinical evaluations of hand function, and for hand rehabilitation devices. Hand biomechanical models have been limited in their ability to accurately predict internal hand forces because adequate force transducers have not been available for measuring individual finger and hand forces during grasping activities, and for determining actual external finger forces applied during different grip configurations and hand functions (An *et al.*, 1985). Hand force sensors are also needed for evaluating the results of surgical procedures such as joint replacement and tendon transfers (Dickson *et al.*, 1972), for force feedback in functional neuromuscular stimulation (Crago *et al.*, 1986), and for sensory substitution rehabilitation devices that compensate for loss of sensation in the hand (Mokshagundam, 1988a).

Conventional methods for measuring finger forces and grip exertions include subjective magnitude estimation (Stevens, 1962), integrated electromyography (Armstrong *et al.*, 1979), and installation of transducers, such as strain gages and force sensors, on tools and objects handled (Fellows and Freivalds, 1989; Radwin *et al.*, 1991). Magnitude estimation, however, is low in resolution and depends on the objectivity of the participant. Surface electromyography is limited to static exertions, is not specific to individual fingers or locations on the hand, and is not practical for measuring forces during complex manual work activities. Force transducers attached to handles must be installed on all objects handled and measurements are limited to the locations the sensors are attached on the handle rather than locations on the hand.

Most commercial force measurement transducers are either too large and bulky for attaching directly to the hands or fingers without significantly degrading manual dexterity, or they are too fragile for withstanding the high forces exerted by the hand. Furthermore, most instruments that have previously been developed for finger and hand force measurement were mostly intended for measuring maximal finger and hand exertions (Dickson *et al.*, 1972; Malinen *et al.*, 1979; Ejeskär and Örtengren, 1981; Bolsinger and Mai, 1985; Jain *et al.*, 1985; Amis, 1987). A force sensor that can be attached directly to the hands can overcome these limitations.

Piezoresistive sensors have been considered for attaching to the fingers but these sensors are highly fragile and brittle (Pennywitt, 1986). Sorab *et al.* (1988) used piezoresistive silicon sensors attached directly to the surgeon's hands for measuring delicate finger forces applied during delivery of newborns. These sensors operated over a force range between 1 and 45 N with a resolution of 0.5 N and had low hysteresis and high repeatability. Their frailty makes these sensors unsuitable for use during rugged manual activities.

Conductive polymer pressure sensors are extremely durable and thin, making them ideally suited for attachment to the fingers and hand. Since they are intrinsically pressure sensors, they must be modified for use as force sensors. This paper presents a method for constructing conductive polymer force sensors, suitable for attaching to the fingers and hands, and for measuring individual finger forces produced during grasp and manual activities.

MATERIALS AND METHODS

A conductive polymer sensing element (Interlink Electronics) was modified for use in the force sensor. These sensing elements are composed of two conducting interdigitated patterns deposited on a thermoplastic sheet facing against another sheet containing a conductive polyetherimide film (see Fig. 1). A spacer placed between the two plastic layers has a cutout that permits the two sheets to make electrical contact when pressure is applied but otherwise causes the sensor to have infinite impedance in the

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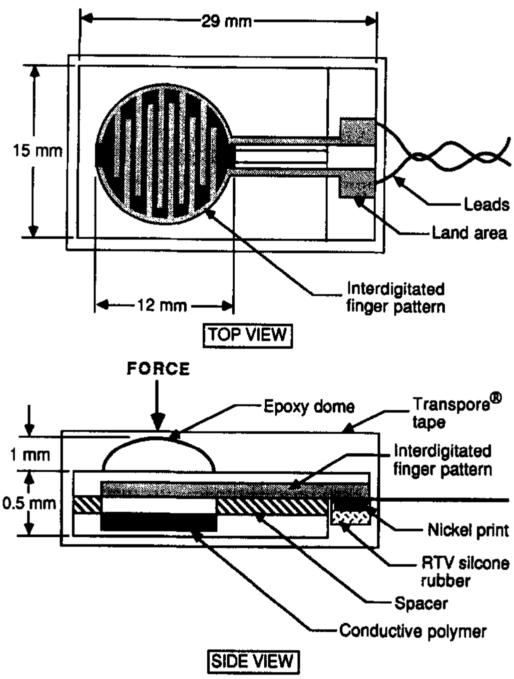

Fig. 1. Schematic diagram of the conductive polymer finger force sensor showing top and side views. After a dome was placed over the conductive polymer sensing area the sensor was encased in Transpore® tape.

unloaded state. As applied pressure increases, the two layers compress together increasing the contact area. This subsequently decreases the electrical resistance and creates a shunt between the interdigitated patterns (Mokshagundam, 1988b).

To create a sensor that responds to force rather than pressure, a method was needed for directing all of the applied force through the effective sensing area. This was accomplished by placing an epoxy dome over the sensing area. The epoxy layer was thin and helped to stiffen the sensing area and keep it rigid under loading.

The sensor selected for measuring finger forces had a 12 mm diameter circular effective sensing area. The overall dimensions of the polymer sheets were $29 \times 15 \times 0.5$ mm. Smaller and larger sensors with square sensing areas (5×5 and 25×25 mm) were tested for comparison and their characteristics are also described in this paper. The effect of temperature on the conductive polymer sensing element was -0.5% full scale per degree Celsius (Maalej *et al.*, 1988).

The first step in constructing the finger force sensor was to expose the land areas so that wire leads could be attached. This was accomplished by removing part of the plastic sheet covering the lands. Wire leads were attached to the lightly sanded lands using nickel print (GC Electronics Model 22–201). The leads were 32–gage, 7–strand wire (Belden 8430 PVC cable). The nickel print was allowed to dry (>24 h), and then a thin layer of RTV silicone rubber (Dow Corning 732 general purpose sealant) was applied over the leads to strengthen the attachment to the land areas.

The surface over the interdigitated patterns was lightly sanded to form a rough surface, and then epoxy (Duro™ Master Meld, Loctite Corporation) was applied in drops over the sensing area until a dome of 1 mm height in the center and a width equal to the sensing area formed. Sensors without the epoxy dome were also constructed for comparison. Transpore® tape (3M Company) was taped around each sensor for protection after the epoxy dried. Microfoam® surgical tape (3M Company) held the sensors on the fingers (see Fig. 2). The total thickness of the completed sensor was 1.8 mm.

Each sensor was placed in the upper leg of a voltage divider circuit using a 6.2 kΩ resistor for the other leg. The output of the voltage divider was amplified using an operational amplifier with a gain of $\times 10$. The resistance of the sensor decreased with increasing force until the resistance reached a minimum. A lower sensor resistance resulted in a higher voltage divider output. The 6.2 kΩ resistor in the lower half of the bridge also served as a current limiter for preventing heating effects in the sensor.

A strain gage dynamometer was constructed for calibrating the sensors. The dynamometer operated by measuring the transverse force acting on an aluminum beam and measuring the strain caused by shearing stress in the cross-section, resulting in an instrument the sensitivity of which was independent of the point of load application (Pronk and Neising, 1981; Radwin *et al.*, 1991). The 161 mm long beam was mounted parallel to a similar beam and the span between the outer surfaces of the two beams was adjustable for a range of 40–130 mm. Aluminum plates (37×116 mm) were attached to both beams for increasing the gripping area.

The strain gage dynamometer was calibrated by suspending weights in the plane of greatest sensitivity. The calibration curve had a slope of $0.040\ \mathrm{N\,mV^{-1}}$ [$F(1, 97) = 148.699$, $p < 0.001$] resulting in a transducer sensitivity of $25 \pm 0.6\ \mathrm{mV\,N^{-1}}$. Output differences for equivalent loads between 1 and 20 N applied at 10 equally spaced points along the length of the dynamometer over a distance of 110 mm were less than 5 mV, which resulted in an error of less than 0.2 N.

A series of tests was performed using these sensors both on the laboratory bench and while attached to the hands of human subjects. Subjects were solicited by posting announcements on bulletin boards in university buildings. Ten subjects participated in this investigation. Four were female and six were male. All 10 subjects were right-handed, and their ages ranged from 18 to 33 yr.

RESULTS

Sensor calibration

Each force sensor was calibrated individually against the strain gage dynamometer by having subjects pinch the dynamometer at 10 equally spaced force levels between 0 and 30 N with the sensors attached to the fingers. Calibration force levels were presented in a random order. Each finger was positioned directly opposite the thumb against the two plates of the dynamometer. A voltage representing a reference force level was subtracted from the dynamometer output signal and the difference was displayed on an oscilloscope screen. The middle of the screen was marked using a small target. After a particular reference force level was established subjects were required to pinch the dynamometer using the finger and thumb until the oscilloscope beam was deflected into the target area indicating that the reference force level was achieved. The experimenter observed the sensor output on the oscilloscope and initiated a data sampling procedure after the dynamometer output remained steady.

The average sensor and dynamometer output voltages were used as static calibration points. The sensor and dynamometer outputs were sampled at a frequency of 40 Hz for a period of 2 s. Summary statistics including the mean, standard deviation and range were displayed after sampling at each calibration level, and the experimenter resampled if the range was greater than 1 N. The 10 calibration points obtained for each sensor were fitted to a second-order polynomial equation using linear regression. This procedure was repeated for each finger. The time necessary for calibrating each sensor was approximately 5 min, including a 30 s rest period between each calibration point.

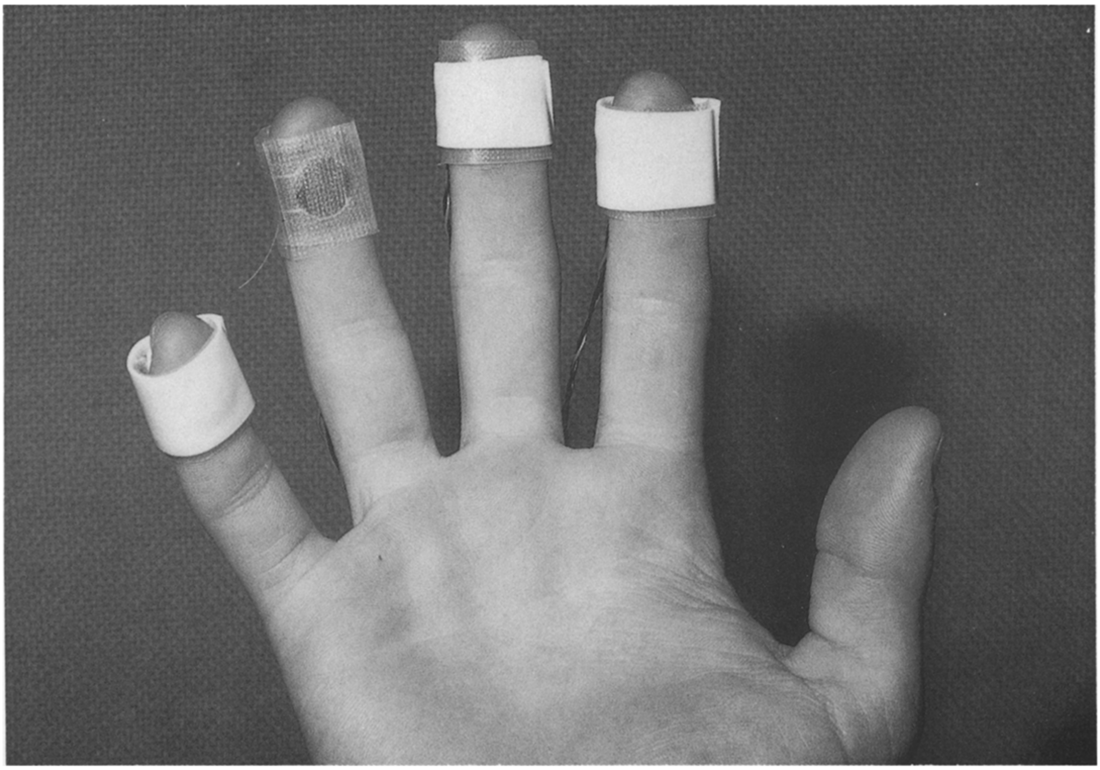

Fig. 2. Finger force sensors were attached to the distal phalangeal pads of the digits using tape. The tape from one digit is removed in this photograph for exposing the sensor.

Least squares regression using a second-order polynomial was deemed adequate [$F(1,7)=20.67$, $p<0.01$] for static calibration of the sensors (see Fig. 3). The average coefficient of determination was 0.98 for 40 replications of this calibration procedure. A third-order polynomial coefficient did not significantly contribute to the calibration variance [$F(1,6)=1.48$, $p>0.05$].

In addition to calibrating the sensors at fixed static force levels, a dynamic method for calibration was also tested. This method was more expedient than the static calibration method, taking less than 30 s per sensor. Subjects were asked to pinch the dynamometer, similar to the static calibration method, except they were instructed to slowly increase the applied force until a 30 N force level was achieved, and then slowly decrease the applied force over a 10 s period. The force sensor and dynamometer voltages were sampled at 40 Hz and every fourth sample was used as a calibration point. A second-order polynomial was fitted similarly to the dynamic calibration data.

The voltage outputs for the static and dynamic calibration for corresponding static force inputs were obtained at 2 N intervals over a 0–30 N range. Linear regression through the origin, using the output for the dynamic calibration as the dependent variable and the output for the static calibration as the independent variable, resulted in an average slope of 0.97 (sensor 1 slope = 0.96, sensor 2 slope = 0.99, sensor 3 slope = 0.96) and an average coefficient of determination of 0.99, indicating the faster dynamic calibration method produced results similar to those using the more time-consuming static method.

Performance as force sensors

A pneumatic cylinder piston rod was positioned over sensors mounted on the finger pad. Sensors both with and without epoxy domes were tested. The force, which was controlled by varying the pneumatic cylinder air pressure, was measured against a strain gage load cell (Interface) in line with the direction of applied force for 5 N intervals over a 0–30 N range. Flat circular styli of varying surface areas (25, 98 and 1100 mm^2) were attached to the end of the load cell for distributing forces over known surface areas. A surface curved convex towards the fingers (16 mm radius of curvature) was also attached to the end of the load cell and applied against the epoxy dome force sensors. The output for sensors both with and without epoxy domes, obtained when applying forces for varying surface areas, is shown in Fig. 4. These data confirmed that the response of the sensors containing epoxy domes was not affected by contact area, while output varied considerably when epoxy domes were omitted.

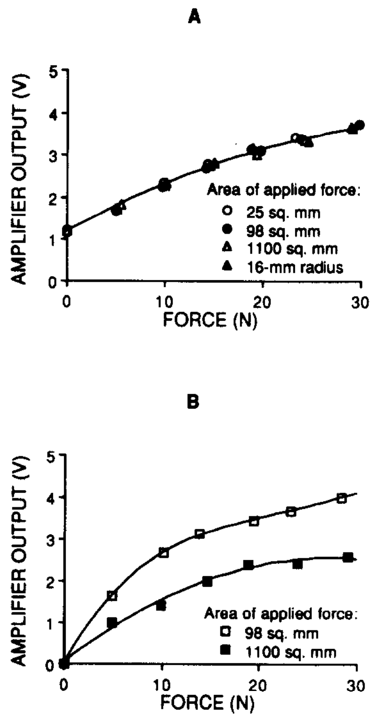

Fig. 4. (A) Response of sensor with an epoxy dome for input forces applied using different size surface areas, including a curved surface, showing insensitivity to area of application. (B) Response of a sensor without an epoxy dome for forces applied using different size surface areas, showing high sensitivity to the contact area of force application when the dome is not included.

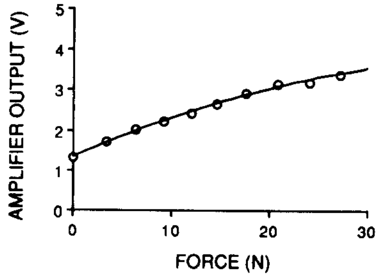

Fig. 3. Typical second-order polynomial regression calibration curve for a sensor indicating the output voltage level for input forces between 0 and 30 N. The d.c. offset voltage at 0 N was variable each time a sensor was mounted on the finger pads making it necessary to recalibrate the sensors every time they were attached.

Shear force loading error

To assess sensor measurement accuracy of compression forces while under shear loading, force measurements were taken at two levels of static pinch force (15 and 25 N), repeated three times for each level while suspending three weights (7, 10 and 15 N) from the dynamometer. Load weights were suspended from a cable attached to the dynamometer that passed through a hole in the table so the load weight remained constant. Force measurements were made while subjects lifted the dynamometer using the index finger and thumb in the same posture used previously. Visual force feedback was used for controlling the actual pinch force level. The average and maximum errors were measured for each of the three weights. Three different sensors had an average error of ± 0.53 N at the 15 N pinch force level and ± 0.61 N at the 25 N pinch force level. The results indicated there was no systematic relationship between shear force and the sensor response. In all cases, the error was less than 1 N.

Sensor step response

Sensor response to a step force input was investigated by placing a 12 mm diameter sensor with an epoxy dome under a pneumatic cylinder piston rod. A solenoid valve controlled activation of the piston in order to obtain a step force input. A strain gage load cell was attached to the piston rod for verifying that the input force rise time was shorter than the sampling interval which was 0.29 ms per sample. Time constants were computed by estimating the time for the sensor output to reach 63% of the final steady state force level using linear interpolation between samples. Sensor time constants were computed for step force inputs of 18 and 29 N for three replications at each force level. The average time constant was 0.57 ms (S.D. = 0.06 ms) for an 18 N force and 0.52 ms (S.D. = 0.13 ms) for a 29 N force. Time constants were measured at two force levels to check for potential non-linearities. No significant time constant differences were observed between force levels [$F(1,4)=0.39$, $p>0.1$] which was expected for a linear system.

Static and dynamic force measurement accuracy

Five similar force sensors were constructed using 12 mm diameter sensing elements and epoxy domes, and were tested for static and dynamic force measurement accuracy. The sensors were each attached to a subject's index finger and calibrated using the static calibration method previously described over three different force ranges (0–30 N, 0–40 N and 0–50 N). The subject was instructed to pinch the strain gage dynamometer at 5 N increments for each of the three ranges and measurements were replicated three times at each static force level for every sensor. To determine the sensor response to slow dynamic loading, forces were applied against the strain gage dynamometer using the fingers while voluntarily varying the exertion force at a rate of approximately one pinch every 2 s for a period of 10 s. Since sensor accuracy was believed to degrade as force level increased, this test was replicated over three force ranges including 0–30 N, 0–40 N and 0–50 N. Representative dynamic responses of the strain gage dynamometer and corresponding conductive polymer sensor output are shown in Fig. 5, indicating good agreement. Lags between the sensor and dynamometer output were negligible.

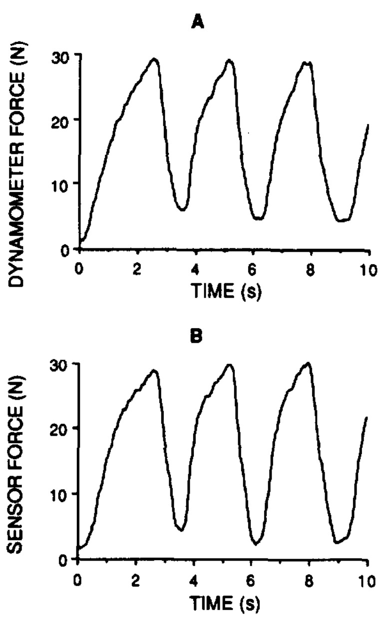

Fig. 5. Force plotted against time to illustrate sensor dynamic accuracy using (A) a strain gage dynamometer as a reference and (B) a conductive polymer sensor with an epoxy dome.

The total error at each force level was computed as the difference between the force measured using the strain gage dynamometer and the force obtained using the sensor output. Table 1 shows the average, minimum, maximum, and the S.D. of the measured error during static and dynamic force measurements for five sensors over each of the three force ranges tested. As the force range increased, average error tended to increase [$F(2,2)=11.92$, $p<0.08$]. No significant differences were observed between average error for dynamic loading and static loading [$F(1,2)=5.14$, $p>0.1$].

Three sensors were also constructed using smaller sensing elements (5 x 5 mm) using the same method described above and they had an average error of ± 1.4 N for a 0–30 N range which was 40% larger than the error obtained for the sensors constructed using the 12 mm diameter sensing elements.

Five-finger pinch force measurement accuracy

Four sensors were attached to the distal phalangeal finger pads of 10 subjects. The sensors were each calibrated using the static calibration method described previously. Seated subjects were instructed to pinch the suspended dynamometer using all four fingers of the dominant hand in a pulp–pinch posture against one plate as the thumb opposed the fingers on the opposite plate. The dynamometer was grasped so that the parallel plates were held vertical. The subjects were provided visual feedback of their total pinch exertion levels, measured by the dynamometer, using the oscilloscope display.

Individual finger forces were measured using the sensors against the dynamometer during five-finger prehension for three force levels and two pinch spans. The force levels were 10, 20 and 30% of maximal pinch strength for each subject, and the spans were 45 and 65 mm. Each combination of force level and pinch span was repeated three times in random order. The subjects also lifted the dynamometer from a table until it was elevated at a height corresponding to a 90° elbow angle, with the upper arm parallel to the upper torso and the wrist deviated so that the parallel plates were held vertical. Total pinch force and individual finger forces were measured while lifting the dynamometer using the three weight levels and two pinch spans. The weights were 10, 15 and 20 N and the two spans were 45 and 65 mm. Each combination of force level and span was repeated three times in a random order. Subjects were instructed to use a grip that they felt they could maintain for a long period of time. After the dynamometer was lifted and a steady grip force was achieved, the dynamometer and sensor outputs were sampled.

Average total pinch force increased from 11 N at the 10% maximal voluntary contraction level to 31 N at the 30% maximal voluntary contraction level. Total pinch force increased from 15 to 31 N when the load weight increased from

Table 1. Errors measured for five sensors using 12 mm diameter sensing elements

Force application	Force range (N)	Average error (N)	Minimum error (N)	Maximum error (N)	Standard deviation (N)
Static	0–30	1.0	0.0	3.2	0.8
	0–40	1.9	0.0	4.6	1.2
	0–50	2.1	0.2	4.4	1.2
Dynamic	0–30	1.2	0.0	3.0	0.7
	0–40	2.4	0.0	5.5	1.2
	0–50	3.2	0.0	6.1	1.5

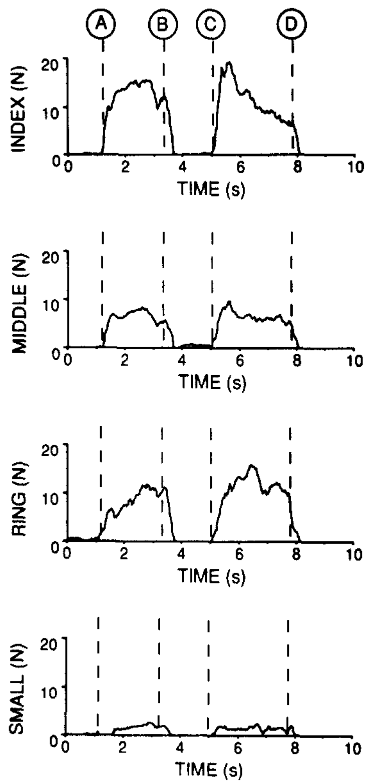

Fig. 6. The finger forces measured during a box-lifting task. (A) The box was lifted from the floor. (B) The box was placed on a table. (C) The box was lifted from the table. (D) The box was placed back on the floor.

10 to 20 N. When summing the forces measured from all four fingers and comparing the sum with the force measured using the strain gage dynamometer, the average error was ± 3.2 N (S.D. = 2.8 N) for the controlled pinch force trials and ± 4.8 N (S.D. = 3.7 N) for the load lifting trails.

Force measurements for typical pinching tasks

The sensor capabilities were tested for measuring forces produced while performing simple manual activities. Forces applied by the fingertips were measured for two representative carrying and lifting tasks. The sensors were calibrated using the dynamic calibration method described.

A subject lifted an 83 N wooden box from the floor and onto a table with the hands pressing against opposite sides of the box. A record of the finger forces measured during the box-lifting task is shown in Fig. 6. Similar measurements were made while performing a book-shelving task consisting of grasping a 19 N book from a shelf and moving it to another shelf. All finger forces measured were less than 20 N for the lifting task and less than 10 N for the book-shelving task, which were within the useful operating range for these sensors.

DISCUSSION

The objective of this study was to investigate a force sensor capable of being attached to the palmar surface of the fingers and hands. Although the manufacturer calls these conductive polymer sensing elements 'force sensing resistors', this was found only in the case when forces were applied using a constant surface area. The conductive polymer sensing element by itself is not a true force sensor. Because it is sensitive to pressure, it can only make repeatable force measurements when the applied force has a constant shape and force distribution. Furthermore, these sensors behave as true pressure sensors only when the area of the applied force is as large, or greater than the active sensing area (Pax *et al.*, 1989). To overcome these limitations, many of the commercially available force sensors used for robotic end effectors have some protruding shape, such as a plastic wedge, to concentrate the applied force over the sensing element (Rittler, 1988; Pennywitt, 1986). A function of the epoxy dome geometry was to ensure that the sensor made contact with only one location during grasp. Without the epoxy dome the sensors did not respond repeatably for the same force applied using different surface areas (see Fig. 4). The data demonstrate clearly that after attaching an epoxy dome the sensors acted as true force sensors and they responded predictably for a given force input even when the contact surface area and curvature were varied.

The conductive polymer sensing elements were sensitive to bending. The sensors responded differently when attached to a finger than when placed on a flat surface, apparently due to the bending incurred when conforming to the finger shape and the pressure applied by the surgical tape to hold it in place. The rigid epoxy domes were added for helping to prevent the sensors from bending under load. After the epoxy domes were attached the sensors were relatively rigid, but they were able to bend enough to conform to the shape of the finger when taped to the hand. As a consequence of the necessary bending, variable offset voltages resulted in the unloaded state each time the sensors were attached to the finger (see Fig. 3). Bending was apparently not a problem for the load range between 0 and 30 N, as evidenced by the relatively low errors (<1 N) measured. Bending may have been a contributor, however, to the somewhat larger errors produced for loads greater than 30 N.

Constant compression of the sensors when the tape was wrapped around the finger was impossible to control. The actual offset voltage when a sensor was attached to the finger was highly variable and depended on the degree of bending of the sensor and the tightness of the surgical tape. There was also a large amount of variability (20%) in the offset voltage between individual conductive polymer sensing elements. Therefore, it was concluded that for reliable results each epoxy dome force sensor had to be individually calibrated on the finger every time it was attached. Because of the small time-constant, either static or dynamic calibration methods produced calibration curves that corresponded well with each other and, therefore, either of the two methods may be used for calibrating the sensors.

Crago *et al.* (1986) recommended that an adequate sensor for evaluation of hand function should have an operating range between 0.1 and 80 N, a resolution of 0.1 N for light touch, and a resolution of 1 N at higher forces. Furthermore, they recommended that the sensors have low mass, small size, ease of mounting, and avoid limiting the range of movement or degrees of freedom. The sensors should also have a minimal number of wire leads to avoid cable bulkiness and wire breakage and they should work well for a wide variety of contact surfaces and operate in changing environments. The sensors described in this paper satisfy most of these criteria, except for the force range.

The force range of these sensors is limited between 0 and 30 N for a resolution of 1 N. This force range, however, is adequate for measuring forces produced by the fingers during the representative manual tasks tested, although the range is not large enough for measuring the forces exerted by the thumb (up to 44 N). Since the sensor response characteristics were quadratic the sensitivity of the sensors decreases as applied forces increase (see Fig. 3). Higher force ranges may be measured using these sensors, but their resolution decreases due to the lower sensitivity of the sensors at higher force levels (see Table 1). This is because as the slope of the calibration curve decreases, the same noise level at higher force levels spans a greater range of forces.

The conductive polymer sensing elements having a 12 mm diameter circular area were selected for use in the finger force sensor because they were small, flexible, and easily conformed to the curvature and various sizes of the fingers. The larger sensors (25×25 mm square sensing area) were not used because their large size impaired dexterity and did not easily conform to the shape of the finger. The smallest sensors tested (5×5 mm square sensing area) were not used because they produced 40% greater average errors (± 1.4 N) over the 0–30 N range than the 12 mm diameter sensors.

These sensors had low sensitivity to shear forces. The average errors (± 0.53 N at the 15 N pinch force level, ± 0.61 N at the 25 N pinch force level) when weights were suspended from the dynamometer were not greater than those measured during the force measurements made while the dynamometer was suspended in order to eliminate shear (see Table 1). The sensors were mostly sensitive to compressive forces since their resistance was an inverse function of compression of the conductive polymer, resulting in increased contact area and hence increased shunting between the interdigitated patterns when forces were applied perpendicular to the patterns and conductive polymer. Shearing forces were likely to have had less affect on the resistance because the contact area between the interdigitated pattern did not change when forces were applied tangentially.

These force sensors were fundamentally useful for measuring pinch exertions typically performed at low frequency loading with little appreciable error. Considering that the sensor mass is low and the stiffness high, a first-order instrument may be a practical approximation. The average time-constant for these sensors was 0.54 ms which then results in a cutoff frequency of 1850 Hz. This bandwidth should provide sufficient responsiveness to normal physiological loading. Hence, the negligible phase and amplitude error (see Fig. 5) is an expected result for an input frequency of 0.5 Hz and a cutoff frequency of almost 2 kHz. Since these sensors may contain nonlinearities such as hysteresis (Pax *et al.*, 1989), future studies should be performed to model more completely the sensor dynamic response.

Despite the limitations of conductive polymer sensors there are few alternatives available for measuring external finger forces. Several practical uses for these sensors (i.e. measuring finger forces produced during carrying and gripping activities) have been demonstrated which are representative of numerous applications in ergonomics and hand rehabilitation. The forces exerted during these tasks were within the operating range (<30 N) of these sensors (see Fig. 6). It is anticipated that these sensors will be useful for numerous biomechanics applications both inside the laboratory and out in the field.

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