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Wearable strain gauge-based technology measures manual tactile forces during the activities of daily living

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Abstract

Introduction: Current methods of determining applied forces in the hand rely on grip dynamometers or force-measurement gloves which are limited in their ability to isolate individual finger forces and interfere with the sense of touch. The objective of this study was to develop an improved force measurement system that could be used during various activities of daily living.

Methods: Custom-made strain gauge sensors were secured to the fingernail of four fingers and two middle phalanges and calibrated to measure hand forces in eight healthy individuals during five activities of daily living.

Results: These sensors were capable of measuring forces as small as 0.17 N and did not saturate at high force tasks around 15 N, which is within the envelope of forces experienced during daily life. Preliminary data demonstrate the ability of these tactile sensors to reliably distinguish which fingers/segments were used in various tasks.

Conclusions: Until now, there has been no method for real-time unobtrusive monitoring of force exposure during the tasks of daily life. The system used in this study provides a new type of low-cost wearable technology to monitor forces in the hands without interfering with the contact surface of the hand.

Keywords

Activities of daily living, tactile sensors, wearable technology, impairment, biomedical devices, patient behaviour monitoring devices, sensor design, sensors/ sensor applications

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Introduction

The human hand is our primary tool used to interact with the environment around us.[1] As a result of the kinematic structure of the upper extremity, our hands and fingers have a high degree of dexterity and are capable of performing a variety of fine motor movements which allow us to perform activities of daily living (ADL).[2,3] Impairment to the fingers as a result of trauma, autoimmune diseases, and degenerative diseases greatly impedes our ability to perform functional tasks.[3] In addition to a reduction in dexterity of the fingers, these impairments often result in pain whenever force is applied to the hands.

Current methods for determining forces in the hands typically involve either a dynamometer or some variation of a force glove.[1,4–6] While dynamometers provide a highly repeatable and accurate measure of hand force,[4,5] they do not allow for force measurement of individual fingers or finger segments. Additionally, dynamometers are unable to measure force during a functional task.[5] Some sensorized glove constructs can measure forces in different finger segments and

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can be used during the tasks of daily living.[5–7] However, sensor gloves occlude the surface of the volar dermis and do not allow for natural tactile feedback during the activities.[5–7] To solve the issues presented by dynamometry and force glove-based measuring systems, some researchers have embedded force transducers into devices that represent some common tasks.[8–10] While these devices are able to measure individual finger forces and simulate a small number of ADL's, they are costly, only crudely resemble ADL's, and cannot be used to measure forces during the actual performance of daily activities.

We propose an alternative that addresses these issues in current touch force measuring systems. This study examines the use of strain-gauges applied to the finger nails and middle phalanges, which detect strains from tissue deformations that occur during contact with objects in common tasks of daily living. The main segments of the finger that are being targeted with this new strain gauge technology are the three joints in the fingers and the two in the thumb, which are the metacarpophalangeal (MCP) joint, proximal interphalangeal (PIP) joint, and the distal interphalangeal (DIP) joint.[11] By securing the strain gauge sensors to the fingernails and on the middle phalanges, measurements of the forces acting at these different locations of the fingers can be made. The two main components of the fingertip that are of interest are the palmar surface, known as the fingertip pad, and the dorsal surface, which is the fingernail (or nail plate). Specialized sensory neurons of the fingertip pad (e.g. Pacinian and Meissner corpuscles, and Merkel cells) differentiate between different sensory conditions such as light and firm touch, temperature and pressure changes, and the differentiation of textures.[12] The fingernail on the dorsal surface of the fingertip is composed of hard, keratinized proteins that protect, provide thermoregulation, and provide tactile sensation, by acting as a counterforce to the fingertip pad.[13] Proximal to the nail plate is the eponychium soft tissue, which sits superior to the distal terminal phalanx of the finger and consists of a network of mechanoreceptors that sense changes in nail curvature and force direction transmitted from the fingertip pad.[14] The application of strain gauges on the dorsal aspect is intended to take advantage of this physiological process for the purpose of measuring finger pad contact forces.

While the concept of applying a strain gauge to the dorsal aspect of the finger is straight forward, it is not clear whether such a strain gauge construct can be sensitive enough to detect tissue strains during low-force functional tasks, and yet not get saturated during high-force tasks. Additionally, the efficacy of such a construct for daily tasks has not been reported. Therefore, the objective of this study was to develop a portable measuring system that would be able to measure the forces occurring in the fingers without interfering with the natural function of the hand grip interface, and not occupying the volar finger pads.

Methods

Tactile sensors

Tactile sensors were created to be placed on four of the fingernails (one sensor on each nail) and two of the middle phalanges (two sensors per phalange) (Figure 1). Each of these tactile sensors consist of one foil type strain gauge (CEA-13-062UW, 350 Ω, Omega Inc., Wendall, NC), which is cemented to a standard acrylic nail substrate commonly used in aesthetic applications (Nailene, Pacific World Corp., Aliso Viejo, CA). Wires were soldered to each of the strain gauges in a three-wire configuration.[15] Sensors were cemented to 16 different sizes of acrylic nails to better match the curvature of the participants' nails, according to the manufacturer's instructions (Omega Inc., Wendall, NC). Once the curvature of each of the participant's nails were determined, a sensor connected to the corresponding size acrylic nail was temporarily affixed to the participant's nail using a double-sided adhesive as per the manufacturer's instructions (Nailene, Pacific World Corp., Aliso Viejo, CA). Each uniaxial strain gauge was placed with its measuring axis oriented transversely across the finger, such that the major bending strain measurement is related to the change in curvature of the nail or dorsal soft tissue during the applied volar force. As force is applied to the finger pad, the bone and tissue below the nail push upward and flatten the curvature of the nail, and similarly for the soft tissue of the middle phalange. This bending strain is measured by the tactile sensors.

Calibration

Each tactile sensor was calibrated to allow for conversion from microstrain to force. The deformable finger tissues themselves form part of the transducer because their stiffness constitutes the relationship between the tactile force and bending strain of the dorsal aspect. In order to account for any physical changes in the finger-gauge construct on each new application, a new calibration relationship is required each time a sensor is mounted to the finger, thus insuring reproducible performance with a single use calibration equation. In order to obtain the calibration, the participant was instructed to press each finger and the two middle phalanges on a clinical finger press load cell (model PF002, NK Upper Extremity Assessment System, NK Biomechanical Corp., Minneapolis, MN), increasing

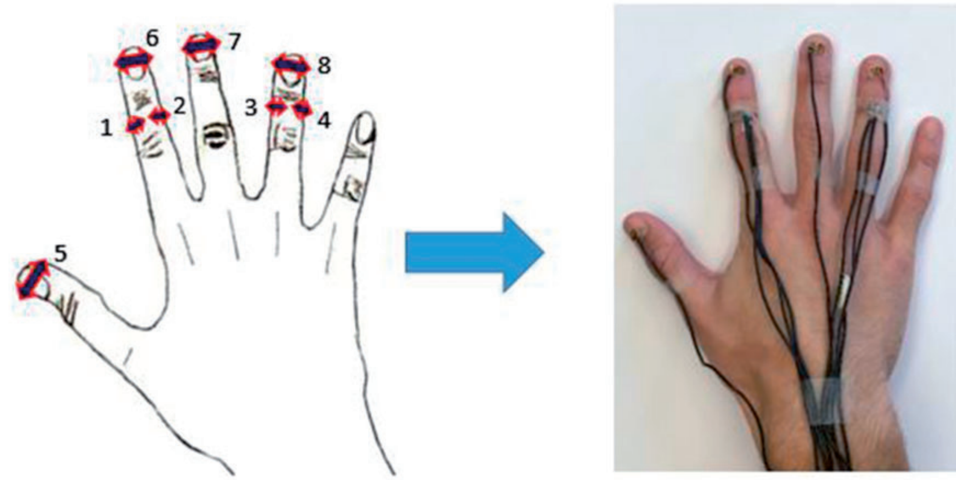

Figure 1. Finger force sensor placement. A total of eight sensors were placed on the dominant hand of each participant. The arrows in the sketch on the left indicate the measurement axis of the sensors. The sensors were placed as follows: (1) Radial aspect of the index middle phalange; (2) Ulnar aspect of the index middle phalange; (3) Radial aspect of the ring finger middle phalange; (4) Ulnar aspect of the ring finger middle phalange; (5) Thumb nail; (6) Index nail; (7) Middle nail; (8) Ring nail.

the force gradually until they reached a force of around 15 N. This force limit was selected as forces beyond this applied to the finger become uncomfortable and outside the range for the tasks being performed, as determined from the pilot tests. Custom written code (LabVIEW, National Instruments, Austin, TX) recorded time-stamped force data from the load cell in Newtons, while the SensorConnect software collected time-stamped microstrain data from the tactile sensor. Force output in Newtons from the load cell was synchronized with microstrain from the finger sensor to produce a calibration equation, which was used subsequently to convert the microstrain data from all trials into the units of force (Figure 2). As seen in Figure 2, a scatter plot was generated for the data collected during the calibration with the microstrain data plotted on the x axis and force on the y axis. A linear trend line was fitted to this data and the resulting equation allowed for the direct conversion of recorded microstrain values during the testing of the force. An example of this calibration being used to convert microstrain to force during an activity can be seen in Figure 3. This process was used for each of the eight sensors and was performed for each individual as the individual's fingers were part of the force transducer.

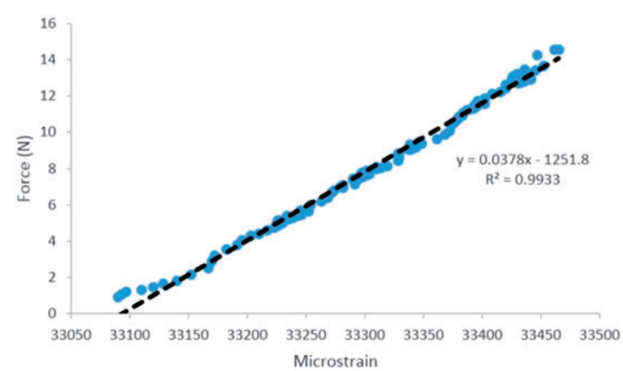

Figure 2. Representative calibration curve. Each sensor was calibrated after being installed on the subject. Subjects were instructed to press each finger instrumented finger segment against a load cell with up to 15 N of force on their volar surface. Force output in Newtons from the load cell was synchronized with microstrain from the finger sensor to produce a calibration equation, which was used subsequently to convert the microstrain data from all trials into units of force. The fit for this sensor was linear with an R^2 value of 0.993.

Study protocol

Eight healthy control subjects (4 female: 21–55 years old, 4 male: 22–52 years old) were recruited for the study. None of these participants reported any pain in their hands or any hand injury. Five representative tasks of daily living involving the hand were examined in this study: zipper, snap button, filling a mug, push button microwave, and open a jar (Figure 4). These tasks were selected from common tasks found in current psychometric evaluations for individuals with hand/wrist pain.[16–18] The first two tasks (zipper and snap button) represent common dressing tasks and are included in both the patient-rated wrist evaluation (PRWE)[16] and the disabilities of the arm, shoulder and hand questionnaire (DASH).[17] The majority of the tasks included in these psychometric evaluations are kitchen tasks. As such, three of the five tasks included in our study are kitchen tasks. Both the jar-opening task and the kettle pouring task were included in the

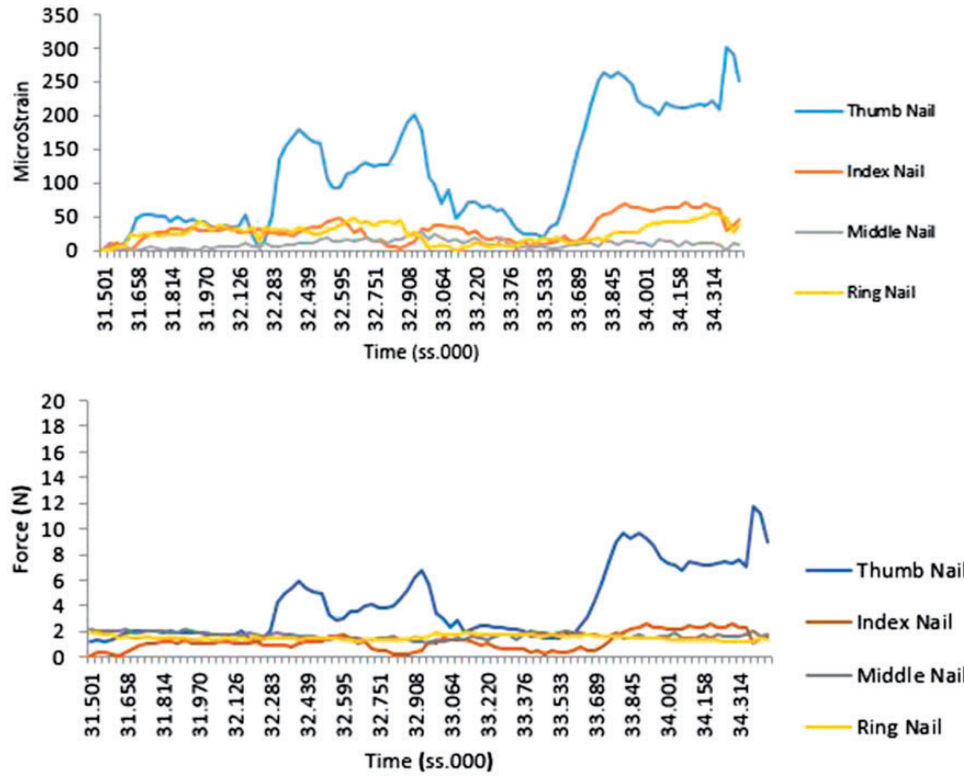

Figure 3. Representative microstrain and force plot. Using each sensor-specific calibration equation (Figure 2), microstrain data recorded during a task were converted to force (N) to allow for meaningful comparisons. This figure illustrates uncalibrated data collected in Microstrain during a zipper pull in the top plot and the same data calibrated to Force (N) in the bottom plot.

joint protection behavior assessment (JPBA).[18] The push button microwave task is not included in current evaluations; however, this task is a ubiquitous kitchen task and thus was included in our protocol. Further, these tasks were chosen to include power and precision grip as well as point loading of individual fingers, which allowed us to examine the sensor's ability to measure these various grips and types of forces. The subjects were asked to perform these tasks three times each and were instructed how to perform the tasks to minimize variability in the data (Figure 4). Including the initial calibration and calibrating between tasks, total testing time was less than 2 h with each maneuver lasting only seconds, thus fatigue was not a concern.

Data collection

Once connected to the participants' fingers, the sensors were connected to a wireless data acquisition transceiver (V-Link LXRS, LORD MicroStrain Inc., Williston, VT) in a quarter-bridge circuit configuration.[19] The transceiver provided signal conditioning and transmitted measured microstrain data wirelessly to a base station (WSDA-Base-104-LXRS, Lord MicroStrain Inc., Williston, VT) connected to a laptop computer running SensorConnect software (Lord MicroStrain Inc., Williston, VT) via USB. Time-stamped microstrain data were sampled at a frequency of 32 Hz, which is more than three times the highest voluntary frequency of human motion of 10 Hz.[20] The wireless data acquisition transceivers were secured to the participant's forearm using Velcro straps and the wires connecting the tactile sensors were secured using tape (Figure 5).

Results

Data were collected for eight individuals during the five ADL examined in this study. Figure 6 shows the average peak force and standard deviation for each of the eight sensors (thumb, index, middle, and ring nail, and each of the two sensors on the middle phalanges of the index and ring fingers) for all the five tasks. The data in Figure 6 indicate that the average maximum force in the fingers is less than 15 N for the activities examined, and was within the expected envelope of force expected for ADL.

Precision grip tasks (zipper and snap button) showed that the highest values recorded were for the thumb and index finger (Figure 6(a) and (b)). For the zipper task, 42% of the overall force measured between the eight sensors occurred in these two fingers and 31% in the

Figure 4. Standardized tasks. The five tasks examined here are: (a) zipping a zipper using the thumb and index finger (precision grip), (b) snapping a button with the thumb and radial side of the index finger (precision grip), (c) opening a jar with the fingertips (power grip), (d) pouring a kettle with four fingers wrapped around the handle and the thumb on top (power/hook grip), and (e) opening a push button microwave using the thumb (point loading).

snap button task. Conversely, for the power grip tasks (opening a jar and filling a mug from a kettle), the highest force values recorded were distributed primarily to the fingertips or the phalanges, depending on the finger segments used to perform the task (Figure 6(c) and (d)). In the pouring task, which utilizes the phalanges to bear the load, 60% of the overall measured force occurred in the phalanges. Conversely, during the jar-opening task which utilizes the fingertips, 62% of the force occurred in the fingertips. Additionally, the task which had a point load in one finger (push button microwave) showed the highest recorded force (20% of the total force) in the thumb, which was used to press (Figure 5(e)). This preliminary data demonstrate the tactile sensors ability to detect strain data from low-force tasks in daily living and not saturate at higher force tasks, while providing information about which fingers/segments were used for each task.

Discussion

This study examined the use of wearable sensors to measure applied loads in the fingers during the ADL. Preliminary results demonstrate the efficacy of these sensors, and early experience indicates that they are sufficiently rugged and reliable for at least a few hours of measurement during some common daily activities. This wireless system allowed the participant

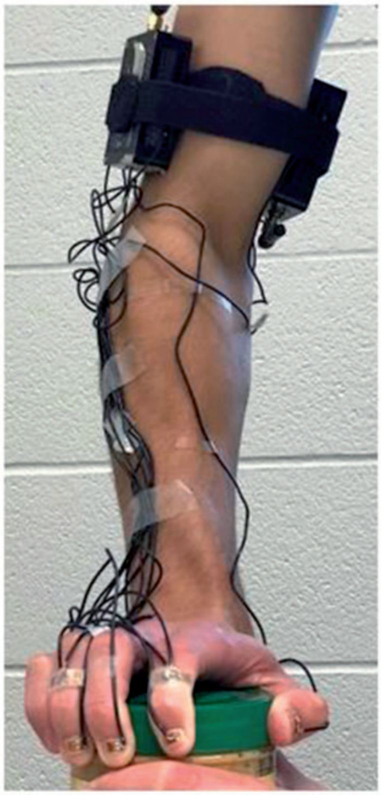

Figure 5. Array of force sensors connected to wireless transceiver. Participant wearing the whole sensor array opening a jar with the wireless transceiver secured to the forearm.

to move about the room freely and perform the activities without being tethered to the computer. The force measured by these sensors was shown to increase as a function of increased applied force throughout the full range of forces expected to be experienced during the normal activity. The nail sensors were capable of measuring to above 15 N during the high-force tasks, where the sensors did not max out or saturate. The ability to measure forces in this envelope suggests that these tactile sensors are well suited for this application.

The results presented in Figure 6 of the average peak forces for individual sensors followed trends that were anticipated based on the finger segments used for the various tasks, as shown in Figure 4. For precision grip tasks, the results showed a higher peak force in the thumb and index fingertips than in other finger segments. Conversely, in grasping tasks, which used a power grip, there was a more even distribution of the forces across the fingers and the forces were greater in the fingertips or phalanges depending on which was used for the task. These expected results indicate that these wearable devices can be used to quantify ADL. However, there was also a considerable amount of strain measured from finger segments that are not central to a particular task. For example, in the pushbutton task, where the thumb was used to push open the microwave door, there is also a relatively large amount of strain registered from the other finger segments (Figure 6(e)). In this case, it appears that while the thumb was used to press the button, the other fingers pressed against each other and the side of the microwave to brace the thumb's movement. As a result, the sensors on these fingers/finger segments registered this strain.

There were limitations. The sensor was intended to quantify external forces from tasks (i.e. extrinsic forces); however, as noted, they were also sensitive to internal (i.e. intrinsic) forces as fingers made contact with each other and with the hand. Tasks like jar opening, which involve all fingers, were not unduly impacted. However, for tasks like pinching, which involve a small number of finger segments, there is a large range of what subjects do with the non-critical fingers. We did not standardize what participants did with those fingers. For example, when opening the microwave with the thumb, some participants made a fist and others did not, which lead to large standard deviations. While intrinsic forces are important, this requires an additional level of interpretation. Also, we cannot be certain that our calibration method can accurately represent forces caused by tissue strains generated intrinsically. This requires further development.

Due to a limit of eight channels on the wireless acquisition unit, we used the following rationale to select specific phalanges. Firstly, the tips of the fingers and thumb were deemed important. Secondly, we felt it was important to instrument the middle phalange of one dexterous finger and one grip finger, with two gauges each, in order to discern any medial-lateral imbalance of grip force on the finger pad. For this, we selected the index finger and the ring finger, as these are prominent dexterous and grip fingers, respectively.[21] The little finger tip was not instrumented because it was not expected to show trends different from the middle finger tip. Similarly, the middle phalanges of the middle and little fingers were not expected to show trends different from their neighbors.[21]

These sensors have addressed a number of issues with previously available measuring devices. Unlike dynamometer-based measuring devices, they are able to measure forces in individual fingers and finger segments. Additionally, they do not occlude the volar dermis of the hand as force gloves do by placing force sensors on the palmar surface of the hand. Finally, unlike sensorized devices with force transducers built into them, our sensors can be worn on the hand to measure forces during the actual performance of various ADL.

These tactile sensors provide a new type of low-cost wearable technology to monitor hand forces during the

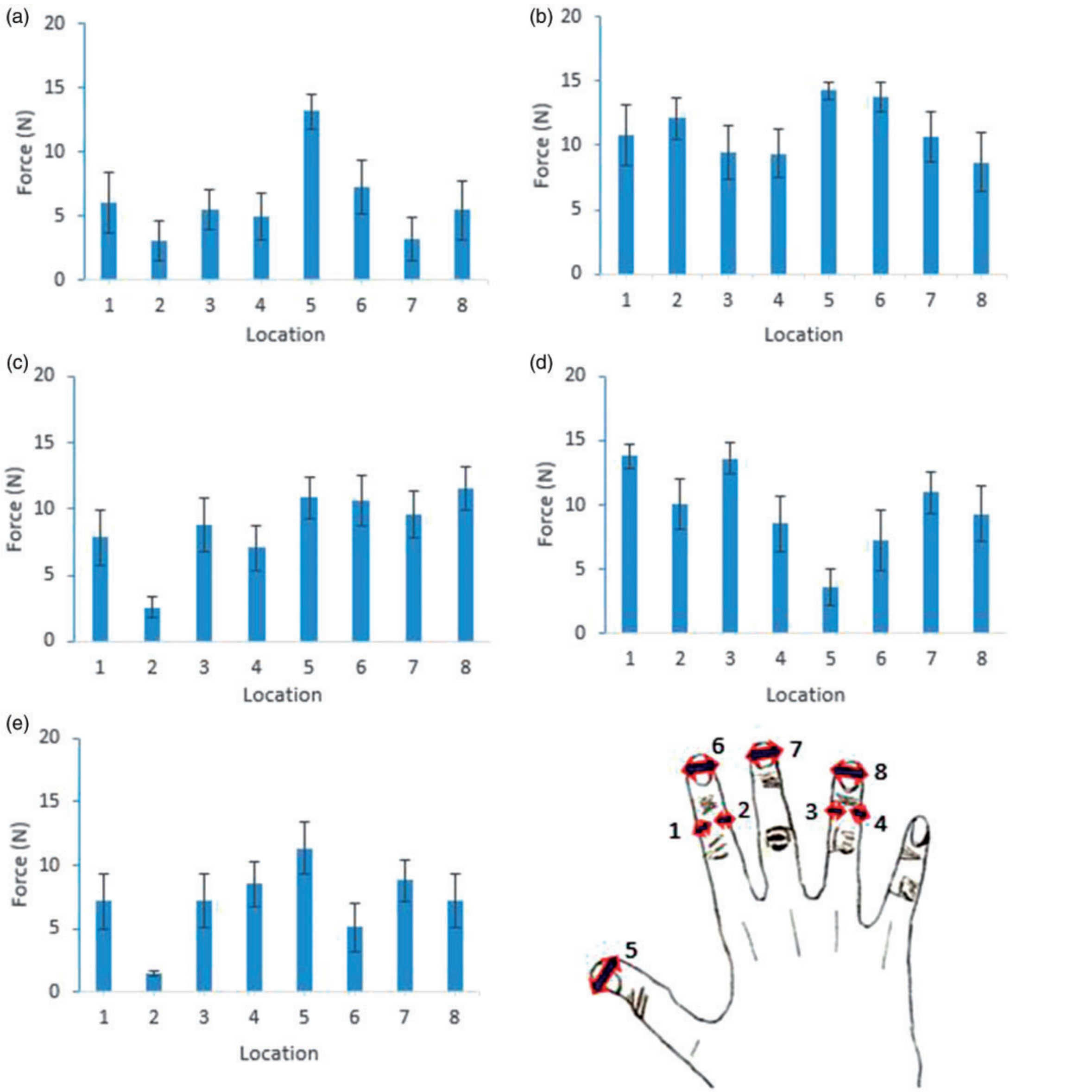

Figure 6. Average peak force for all participants. The peak force for each participant was measured in each sensor for the five tasks performed as defined in Figure 4: (a) zipping a zipper, (b) snapping a button, (c) opening a jar, (d) pouring a kettle, and (e) opening a push button microwave. The average peak force for all participants and standard error is plotted here ($n = 8$ participants) for each of eight sensor locations. For (a) zipping, the force was greatest in the thumb and index finger as expected. For (b) button snap, the force was greatest in the thumb and index finger. For (c) jar opening, the force was greatest in the fingertips. For (d) kettle, the force was greatest in the middle phalanges, and for (e) microwave pushbutton, the force was greatest in the thumb.

ADL that do not interfere with the contact surface between the user and the object. Additionally, their light weight and low profile does not interfere with normal hand function. With this setup of sensors on four of the nails of the dominant hand as well as two sensors on two of the middle phalanges, the forces are distinguishable to each of the fingers as well as finger segments, not just the overall force in the hand. This allows for the determination of tasks which rely more on the fingertips versus the phalanges and information about finger recruitment during various activities. In future applications, more sensors could be added to measure the forces in other finger segments as well.

This investigation reports early experience with a tactile force measurement method, and as such, there were limitations, as should be expected with any new measurement device. Unlike most transducers, in this method, the finger itself forms part of the transducer, because the stiffness of the nail and finger tissues is what constitute the relationship between the tactile force and tissue deformation, which is measured as bending strain on the dorsal aspect. Thus, a calibration is required each time a sensor is mounted to the finger, and so a new calibration equation is determined each time. This may be seen as a limitation, compared to typical hand force sensors such as grip dynamometers. Additionally, the early prototype nature of the current device required that a thin cable for each sensor be routed up the arm to the wireless transmitter. While the finger pads were not occluded, and the wearer was free to walk around untethered, we cannot say with certainty that these extra wires did not influence the wearer.

Improvement of the calibration method is required in order for these sensors to provide a reliable means of quantifying finger contact forces during the functional tasks. This will allow future investigations to evaluate the sensor's efficacy to provide clinicians with objective clinical data in order to inform treatment plans for patients with degenerative diseases, such as osteoarthritis. This system also has the ability of continuous data logging to allow for extended wearability. Future studies are required to monitor patient's activities over an extended period of time in order to tailor treatment plans and provide alternative strategies for daily tasks, depending on the patient's severity and type of degenerative disease, and what tasks they normally perform. When tailoring the system to the patient, the user could receive real-time auditory/visual feedback when a set of threshold force is exceeded to assist the wearer in determining how much force they are using while performing certain tasks, and to encourage them to use modify their behavior to reduce cumulative joint forces. This is a subject of future investigations.

Further investigation should also explore the viability of these sensors to examine ergonomics (both techniques and tools) in the workplace. Factory workers, construction workers, plumbers, and shop operators are jobs which require hand use for extended periods of time while performing strenuous tasks. The wireless system used to transmit measurements in this study has a range of 2 km, and thus a future workplace ergonomics study could allow workers to move throughout the workplace freely while wearing the sensors, and having the data sent to a centrally located computer.

This study has shown that a construct of strain gauges applied to the dorsal aspect of the fingers can measure tactile forces. This simple yet effective method will find several applications in quantifying touch and dexterity.

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Guarantor

LF.

Contributorship

MR analyzed the data, researched the literature, and co-wrote the manuscript. KM collected the data and analyzed the results. SH and EL gained ethical approval, and recruited subjects. EL and LF co-wrote the manuscript and researched the literature. All authors reviewed and edited the manuscript and approved the final version of the manuscript.

Research ethics

This study was approved by the Health Science Research Ethics Board at Western University (protocol ID no. 108625).

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