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New Device for Intrinsic Hand Muscle Strength Measurement: An Alternative to Strain Gauge Handheld Dynamometer

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Abstract

An accurate measurement of intrinsic hand muscle strength (IHMS) is required by clinicians for effective clinical decision-making, diagnosis of certain diseases, and evaluation of the outcome of treatment. In practice, the clinicians use Intrins-o-meter and Rotterdam Intrinsic Hand Myometer for IHMS measurement. These are quite bulky, expensive, and possess poor interobserver reliability (37–52%) and sensitivity. The purpose of this study was to develop an alternative lightweight, accurate, cost-effective force measurement device with a simple electronic circuit and test its suitability for IHMS measurement. The device was constructed with ketjenblack/deproteinized natural rubber sensor, 1-MΩ potential divider, and Arduino Uno through the custom-written software. Then, the device was calibrated and tested for accuracy and repeatability within the force range of finger muscles (100 N). The 95% limit of agreement in accuracy from -1.95 N to 2.06 N for 10 to 100 N applied load and repeatability coefficient of ± 1.91 N or 6.2% was achieved. Furthermore, the expenditure for the device construction was around US\$ 53. For a practical demonstration, the device was tested among 16 participants for isometric strength measurement of the ulnar abductor and dorsal interossei. The results revealed that the performance of the device was suitable for IHMS measurement.

Keywords

force sensor, intrinsic muscles, muscle strength evaluation, cost effective, device

Introduction

Muscle strength refers to the amount of force that a muscle can generate against resistance in a single maximal effort. Evaluation of muscle strength, including intrinsic hand muscles, is frequently carried out by physicians, physical therapists, and other professionals in clinical settings for clinical decision-making and outcome measurements (Schreuders, Selles, Roebroeck, & Stam, 2006). However, the purpose of muscle strength testing varies; it includes prediction of the risk of developing certain diseases in the future (Kim, Kim, Seo, & Kang, 2014; Rikkonen et al., 2012; Timpka, Petersson, Zhou, & Englund, 2014), assistance in the diagnosis of medical conditions (Cruz-Jentoft et al., 2010; Schreuders, Roebroeck, Jaquet, Hovius, & Stam, 2004), examination and comparison of the efficiency of treatment, documentation and monitoring of the progress of muscle strength

during treatment, and providing feedback during the phases of rehabilitation (Rosen, Dahlin, & Lundborg, 2000; Schreuders et al., 2004, 2006). Strength testing is also done among healthy subjects, particularly in sports medicine and ergonomic industries to prescribe suitable strength training.

In this context, intrinsic hand muscle strength (IHMS) evaluation is important for researchers and clinicians, as it determines a human's capability of accomplishing tasks of adaptation, exploration, prehension, manipulation, and perception (Simone et al., 2007). Reduced IHMS can lead to a severe loss of hand function; for example, firmly holding a key will be impossible with the paralysis of the intrinsic hand muscles. Furthermore, atrophy of these muscles alerts clinicians to medical diseases such as leprosy, poliomyelitis, Guillain–Barre syndrome, amyotrophic lateral sclerosis, rheumatoid arthritis, hand-arm vibration syndrome, and diabetes (Videler, Beelen, Aufdemkampe, de Groot, & Van Leemputte, 2002; Vinci, Esposito, Perelli, Antenor, & Thomas, 2003). A few researchers have reported that increased IHMS improves the functional status of the patient and recommended that the method/device used for evaluation of IHMS should be accurate and reliable (Cortez et al., 2011; Hsu, Tang, & Jan, 2000). In clinical practice, IHMS is frequently assessed by the Medical Research Council (MRC) scale, Intrins-o-meter, and Rotterdam Intrinsic Hand Myometer (RIHM; Pataky, Savescu, Latash, & Zatsiorsky, 2007; Xu et al., 2010).

The MRC scale classifies IHMS into six cardinal grades (0–5 muscle grade) based on observation, palpation, gravity, and manual resistance provided by the therapist. In the classification of the muscle strength grade, this cardinal scale places each muscle group into a particular grade that best suits a description. Although the MRC scale facilitates quick clinical evaluation and description of IHMS, it is not much reliable or sensitive in detecting small changes in the muscle strength grades of 4–5 (Harlinger, Blalock, & Merritt, 2015; Schreuders et al., 2000). To overcome the drawbacks of the MRC scale, researchers have developed modified dynamometers, namely, the Intrins-o-meter (Mannerfelt, 1997) and the RIHM (Schreuders et al., 2000). Both these devices are widely used by health professionals in clinical settings to measure the isometric strength of intrinsic muscles. Pataky et al. developed a device based on the model of multicomponent force transducers with complex electronic circuits and claimed that the abductor and adductor muscle strength of all the fingers can be measured instantly by it. However, the available dynamometers are expensive (cost ranges from US$1,000 to US$1,600), quite bulky, and exhibit poor interobserver reliability (37–52%) and sensitivity (Pataky et al., 2007;

Xu et al., 2010). Xu et al. noted that the device developed by Pataky et al. is not suitable for the measurement of IHMS among young children and patients with hand deformities due to its rigid design. Therefore, there is a need for the development of an alternative low-cost, lightweight, and accurate device for the measurement of IHMS.

The working principle of all these dynamometers is based on the piezo-resistive load cell, in which the resistance of the metallic foil strain gauge is altered upon the perpendicular application of compressive or tensile load. The resistance change in the strain gauge is accurately measured using the Wheatstone bridge, and its output is displayed in voltage. The output voltage is then calibrated with the known standard weights and further displayed as a force (in N or kg or lbs) in the device (Xu et al., 2010). Recently, our research group developed a highly reliable, high-pressure sensitive ketjenblack (KB) reinforced deproteinized natural rubber (DPNR) composite (KB/DPNR), equivalent in its behavior to a metallic foil strain gauge sensor. We have successfully monitored grade I and grade II joint mobilization techniques and the lateral pinch strength using KB/DPNR in our earlier study (Madhanagopal et al., 2017). In this study, (1) a new, cost-effective device was constructed, based on the KB/DPNR force sensor, using simple electronic components; (2) the accuracy and repeatability of the developed device were tested with standard applied loads (1–10 kg); and (3) the suitability of the device for IHMS measurement was examined.

Methods

Participants

In this study, the device was tested among 16 participants (4 males and 4 females in each group) from two different age-groups (age range = 30–39 and 60–69). The male participants' mean age, weight, and height in the 30–39 age-group were 34.50 years ($SD = 1.70$), 80.50 kg ($SD = 10.90$), and 1.64 m ($SD = 0.08$), respectively, while the female participants' mean age, weight, and height were 35.80 years ($SD = 4.80$), 62.50 kg ($SD = 7.20$), and 1.57 m ($SD = 0.03$), respectively. The mean age, weight, and height of the male participants in the 60–69 age-group were 64.80 years ($SD = 3.90$), 72.20 kg ($SD = 3.10$), and 1.62 m ($SD = 0.09$), respectively, and those of the female participants were 64.80 years ($SD = 2.80$), 62.20 kg ($SD = 7.50$), and 1.54 m ($SD = 0.07$), respectively. The participants were recruited from a residential apartment and a public university in Johor, Malaysia, through convenience sampling. The inclusion criteria were as follows: participants

had no soft tissue injuries in the hand, no median and ulnar nerve damage, no arthritis of the hand, and were able to understand the instruction provided by the physiotherapist. The participants who did not meet the abovementioned inclusion criteria were excluded from the study. Written and informed consent was obtained from all the participants. The study was approved by AIMST University Human and Animal Ethical Committee (Ref: AUHAEC/EXT/2017/01).

KB/DPNR Sensor Construction

The KB (18 g), each 3 g of cross-linking agents (trimethylolpropane trimethacrylate (SR 350) and dicumyl peroxide), and 300 g of DPNR were hand mixed for 5 min. The hand mixed compounds were then incorporated into a two-roll mill machine and rotated 10 times at a speed of 200 rpm. Finally, the roll-milled mixture was molded under 3,000 psi at 160°C for 10 min and dried at room temperature to obtain a KB/DPNR nanocomposite sheet. A portion of the KB/DPNR (25 mm × 25 mm × 1 mm) was cut from the sheet and utilized for the sensor construction. Then, two thin copper wires were glued on either end of the same plane of the KB/DPNR composite using a silver paste. Finally, adhesive copper tapes were fixed over it for good contact.

Device Construction

The two copper wires of the KB/DPNR sensor were connected to the A_0 and the ground pin of the Arduino, respectively. A potential divider with 1-MΩ standard resistor was adopted, as it exhibited an excellent voltage output range by the KB/DPNR sensor, upon applying compressive load under 5 V constant power supply. For accurate force measurement, the device was first conditioned with a known weight of 11 kg ($\sim$110 N) placed on the sensor. Then, the device was calibrated with the known weights of 1–10 kg with an increment of 1 kg maintained for 5 seconds (Madhanagopal et al., 2017; Tuttle & Jacuinde, 2011). This calibration step was repeated 5 times. In our case, the aim of the developed device was to measure IHMS, and therefore, the device was calibrated up to 100 N, as the finger muscle can generate a maximum of 108.7 N (Schreuders et al., 2000). The mean value of the voltage change to an applied load was taken as the response of the device. To compute the output voltage into a force value, the slope and the intercept values were obtained from the linear equation ($f = y0 + a(x)$), by plotting the mean peak point of the output voltage (V) against the applied

load (kg). In the linear equation, f is the output voltage (V), x is the applied load, $y0$ is the intercept between the output voltage (V) and applied load (kg), and "a" is the slope. The KB/DPNR sensor was glued to the acrylic block (2.5 × 2.5 cm) and mounted to the center of the plastic casing for smooth execution of the muscle strength testing. The device readings were displayed on a liquid crystal display (LCD) using custom-written Arduino software. Finally, the sensor, Arduino, and the LCD were soldered and compactly packed into a plastic casing (8 × 8 cm).

Measures

The accuracy and repeatability of the device were measured by placing the known standard weights of 1–10 kg on the sensor, at an interval of 1 kg, 5 times each (Madhanagopal et al., 2017; Tuttle & Jacuinde, 2011). The readings of the device were recorded using the Arduino software program. The isometric muscle strength of the ulnar abductors (UA) and dorsal interossei (DI) were measured as described by Schreuders, Selles, Roebroeck, and Stam (2006) and Schreuders et al. (2000). In brief, for UA strength measurement, the participants were asked to position their palm facing upward, with the little finger in slight contact with the device. For DI strength measurement, the participant's palm was facing downward with the index finger in slight contact with the device and the remaining fingers stabilized by a physiotherapist. The participants were then asked to generate a force by gradually using their index finger for the DI and their little finger for the UA against the device for 1–2 s and maintain the maximal effort for 4–6 s, respectively. This procedure was repeated 3 times at intervals of 1 min. A 1-min rest was given between repetitions to minimize muscle fatigue. The mean average of the three measurements was recorded as a participant's isometric muscle strength of DI and UA.

Statistical Analyses

Normality distribution of the obtained data was analyzed using the Shapiro–Wilk (S-W) and skewness and kurtosis statistical test. To assess the accuracy and repeatability of the device, the Bland and Altman plot was constructed to describe the agreement between the mean difference of the actual values of all applied load (10–100 N) and readings of the device (N) as well as the mean difference of the repeated measures against the mean force (N). The upper and lower 95% limit of agreement was calculated in

Newton and percentage values. The acceptable limit of agreement was set at $\pm 10\%$ for both the accuracy and the repeatability of the device.

Results

The linearity and the sensitivity (*S*) of the device were $r = .98$ and $S = 0.21$ V/kg, respectively. A S-W test ($p > .05$) for all applied load (10–100 N) and visual inspection of the normal Q-Q plots showed that the accuracy readings of the device were approximately normally distributed for all applied loads (10–100 N), with skewness *z*-values of 0.13, −0.22, 0.19, 0.89, 0.11, 0.66, −1.65, 0.18, 0.19, and 0.74 and kurtosis *z*-values of −1.46, −0.86, −0.76, −0.63, −0.47, 0.54, 1.19, −1.05, −1.18, and −0.20, at an interval of 10 N, respectively.

Accuracy

The mean (*M*) $\pm$ standard deviation (*SD*) differences and standard error of mean (*SEM*) between the readings of the device and applied loads were 0.05 N $\pm$ 1.03 N and 0.14 (0.6% $\pm$ 3.4% and 0.48%), respectively. The 95% limit of agreement between the mean difference from applied load and the readings of the device was calculated to be 2.06 and −1.95 N (7.33% and −6.15%). The Bland and Altman plot revealed that the mean difference from actual value was closer to zero from 10 to 100 N, with a bias of 0.06 (0.59%) in the accuracy. In terms of the percentage, mean differences from the applied load was decreased upon increasing the applied load from 10 to 100 N. All readings of the device strongly agreed with the actual value and closely distributed around zero with low *SD* (3.44%) and *SEM* (0.48%) values.

Repeatability

The *M* $\pm$ *SD* difference and *SEM* of repeatability of the device were 0.006 $\pm$ 3.20% and 0.45%, respectively. The 95% limit of agreement between the differences from mean value and the mean force was 1.92 and −1.91 N (6.27% and −6.26%). The Bland and Altman analysis revealed that the readings of the device have strongly agreed with each other with a mean difference less than 1.92 N, with a bias in repeatability of 0.004 (0.006%). Similar to the plot of accuracy, the percentage of mean differences of repeatability was decreased closer to zero upon increasing the applied load

from 10 to 100 N, which confirms that the readings of the device are repeatable with the same applied load.

Muscle Strength

The obtained isometric strength of UA and DI of all the 16 subjects were as follows: The $M \pm SD$ of DI muscle strength of the 30–39 years age-group (male: 69.47 $\pm$ 5.97 N, female: 56.15 $\pm$ 4.66 N) were higher when compared to the 60–69 years age-group (male: 50.72 $\pm$ 5.63 N, female: 44.42 $\pm$ 4.32 N). Similarly, the measured UA muscle strength of 30–39 years age-group (male: 42.67 $\pm$ 5.66 N, female: 32.85 $\pm$ 4.47 N) were also higher compared to 60–69 years age-group (male: 33.19 $\pm$ 4.29 N, female: 25.82 $\pm$ 4.32 N).

Discussion

Our device works under the principle of piezo-resistive load cell, similar to the commercially available strain gauge handheld dynamometer (HHD). When the device was subjected to loading and unloading of low compressive load (e.g., 30 N), the reading of the device rose up and returned to the zero reading immediately, whereas, under high compressive load (100 N), the device exhibited similar behavior, but it took time (less than 15 s) to return to the zero reading after unloading the applied load. The delay in getting back to the zero reading was due to the minimal hysteresis of the KB/DPNR sensor. However, this hysteresis might not affect the consecutive muscle strength measurement as 30 s–2 min rest is usually given to subjects between the trials of strength testing (Hebert et al., 2011; Symons, Vandervoort, Rice, Overend, & Marsh, 2005).

The device can be used in two routes: as a secured laptop device or a freely handheld device. As a secured laptop device, both the readings of the device and the real-time visual feedback via graph can be monitored, and it would allow the tester to carry out the strength testing not beyond 2 m. As a freely handheld device, it will allow the tester to conduct the muscle strength testing in any clinical setting. However, the limitation is that the displayed force readings cannot be saved during such a mode of testing. In order to eliminate this issue in the present version of device, we are currently working on including a data logger into the device to save the recordings. The budget required for the development of the device and the sensor was analyzed. It was estimated to be approximately US$53, which suggests that the device cost is 20–30 times lower than the commercially available

HHD (US$1,000–US$1,600; Stark, Walker, Phillips, Fejer, & Beck, 2011). The accuracy and repeatability of this inexpensive device were similar to the commercially available HHD. In this study, isometric strength measurements of UA and DI were chosen as a model system. The measured muscle strength of UA and DI by our device was lower in older adults when compared to young adults, which confirms that the developed device is suitable for all age-groups. The intra and interobserver reliabilities of the device are under progress. The proposed method to develop a sensor and a force monitoring device may be useful for health professionals in designing a new device for health-care applications, such as monitoring applied force during joint mobilization.

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Declaration of Conflicting Interests

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