TOWARDS COMBINED FORCE AND DISTANCE SENSING USING ONLY OPTICAL SENSORS TO AID IN STROKE REHABILITATION

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Abstract: In this paper, we present a systematic approach to characterize optical sensors for their suitability to measure both distance and force, when pressed against with the tongue. For this task, a custom test station was developed that can simultaneously record both quantities. Seven different sensors were examined (4 analog, 3 digital, with respect to their output values) with force and distance ranging from 0.1 N to 8 N and 0 mm to 30 mm, respectively, and averaged measurements were compared. Distance characteristics were mainly determined by the half-power angle of the light emitter (with wider half-power angles generally being worse). For force measurements, indicators for good performance (i.e., a monotonically increasing sensor output with increasing force) seem to be a wide half-power angle and no height difference between light source and photodetector. Allthough precise performance parameters for force measurements were not fully identified, using optical sensors for both distance and force sensing is generally possible.

1 Introduction

Strokes are the second major cause of death worldwide, with its prevalence greatly increasing with age. The raw number of non-fatal stroke incidents is even higher, as roughly three out of four patients survive [1]. Stroke survivors struggle with various complications, including orofacial impairment [2]. As a direct result, the majority of stroke patients suffer from dysphagia (i.e., difficulties in swallowing) [3] and dysarthria (i.e., the inability to articulate properly) [4]. These symptoms not only tend to not improve without treatment, negatively impacting the patient's quality of life, but are also a major cause for follow-up complications such as aspiration pneumonia, malnutrition and dehydration [3].

Therapy for dysphagia mainly revolves around strengthening and mobilization exercises for the tongue and several studies showed an improved oral-phase swallowing function after treatment [5, 6, 7]. In fact, a tongue pressure below a threshold of 20 kPa could be used as an indicator for dysphagia in post-stroke patients [8]. As such, several devices have been developed that can measure the tongue pressure, e.g., products from swallowsolutions, iopimedical, or developments from [9] and [10]. These devices consist of one or more piezoelectric, capacitive or volumetric pressure sensor(s), mounted on a flexible support that can be held against or placed on the patient's hard palate.

For treatment of dysarthria, speech-related exercises are neccesary to target articulatory deficits and to restore proper speech production in patients [4] as no clinical evidence exists that strengthening exercises alone are sufficient [11]. However, during these exercises, the patient's tongue is not easily visible from the outside and the difficulty for the speech therapist remains to judge whether the patient is performing the exercises correctly. A viable solution to this problem is the use of optical tongue sensing methods as described in [12, 13, 14, 15] which register the tongue movement inside the oral cavity.

To aid both dysphagic *and* dysarthric patients effectively, tongue position *and* pressure need to be measured with the same device and with a reasonable spatial resolution. As space inside the oral cavity is very limited, it is highly beneficial to measure both quantities with the same sensor type. By exploiting the effect of subsurface scattering in human tissue (and consequently the tongue), optical sensors pose a viable solution. Chuang and Wang [13] already noted this effect in their early work on optical distance sensing, but did not make much use of it as they were mainly interested in the tongue's position. Ohmura et al. [16] did make use of subsurface scattering in polyurethane foam to design a pressure sensor, which is however limited to pressure sensing only, as it covered the photo detector. In this paper, we present the systematic approach to select and analyze several different optical sensors for their suitability along with a newly developed measurement system to measure both distance and force.

2 Sensor choice

Suitable sensors have to meet a number of conditions. They need to be small and relatively flat (with dimensions in the range of a few millimeters) with no sharp edges or lenses that could disturb or even injure the patient's tongue. It should be possible to encapsulate the sensors into some form of polymer to protect them from electrical shortages without degrading their optical characteristics or function in general. Furthermore, they should consume little power, be inexpensive, commercially available and active (with respect to their lifecycle). These constraints already heavily limited the range of commercially available sensors. The remaining ones can be separated into two groups with respect to their output signal, i.e., analog and digital sensors.

2.1 Analog sensors

Analog sensors consisted of a light emitter (either a conventional LED or a laser-diode) and a photo-detector (phototransistor or photodiode). Both parts were connected to appropriate circuitry to convert the photocurrent into a voltage that could be read out, e.g., by a microcontroller. Different combinations of light emitter, photoelement and circuitry were tested. These combinations are summarized in Table 1 and the basic circuitry used is shown in Figure 1. The light emitters were driven by an adjustable current source.

Short label	Components	Circuitry
A1	OPR2800V (TT Electr.) + TEMT7100x01 (Vishay)	Fig. 1(a)
A2	OP280V (TT Electr.) + TEMT7100x01	Fig. 1(a)
A3	TCRT1000 (Vishay)	Fig. 1(a)
A4	OPR2800V + VEMD6160x01 (Vishay)	Fig. 1(b)

Table 1 – Different combinations of light emitter and photoelement, sorted by the circuitry used.

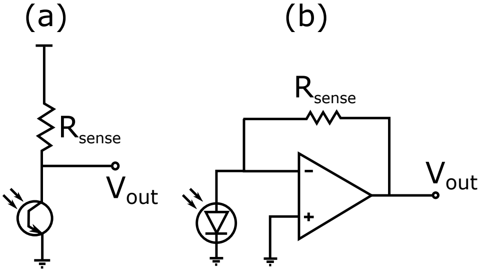

Figure 1 – Basic circuitry used for the conversion of a photocurrent into a voltage.

2.2 Digital sensors

The digital sensors were integrated circuits already equipped with a light source, photodetector, a signal processing chip and ADC. All of them had an I2C-communication interface available to read out the sensor values. On top of that, several registers could be accessed to control the LED current as well as the pulse duration and frequency for different sampling rates. The tested digital sensors are summarized in Table 2.

Short label	Sensor name	ADC resolution
D1	VCNL4010 (Vishay)	16 bit
D2	VCNL4020c (Vishay)	16 bit
D3	SFH7779 (Osram)	12 bit

Table 2 – Tested integrated sensors with digital outputs.

3 Test station and sensor setup

In order to characterize both the sensors' distance and force measuring capabilities, a test station as shown in Figure 2 (left) was designed. Its frame consisted of a base plate along with a headrest and a cantilever, both of which were vertically displaceable along a slide rail. A strain gauge (type: TAL220, Sparkfun) extended the cantilever at the end of which a sensor clip was attached to. Each sensor could be slid into the clip until centered (Figure 2, b) and a spacer (as in [15]) was magnetically attached to the clip (Figure 2 right, (a)), providing a fixed distance between the tongue and the sensor. Figure 3 shows examples of the standardized sensor layout. Every sensor was encapsulated by an artificial gum substrate (GC Reline soft) and cast into silicone to prevent shortages when touched by the tongue and to allow for easy surface disinfection. The headrest and cantilever were adjusted until a comfortable position was found and locked with two small pins. The test station's base plate was firmly screwed to a table using a vice-grip wrench. An Arduino Uno board with its ATmega328 microcontroller was used to interface the strain gauge amplifier (type: HX711 with breakout board by Avia/Sparkfun) and the different sensors. To read out the signal of the analog sensors, the board's 10 bit analog-to-digital converter was used while the values from the digital sensors (varying ADC resolution, see Table 2) could be read out of their internal registers using the I2C-protocol. The measurements were recorded via a serial USB-connection using Matlab and a custom-build GUI. Evaluation and plotting of the sensor values was done with Matlab as well.

4 Measurement and evaluation procedure

The strain gauge was initially calibrated by locking the cantilever into a horizontal position and placing weights (in the range of 10 g to 500 g) at the end and directly above the sensor's position. The force exerted onto the sensor could then be calculated with $F = c \cdot ADC \cdot g$, where c was the slope [g/LSB] found by calibration, ADC the digitized light signal and g the acceleration of gravity. The measurement sampling rate was capped at around 12 Hz as the strain gauge amplifier would not permit higher sampling rates without physically modifying it. The light emitter currents were set to 7 mA for sensor A1, A2 & A4, and 40 mA for sensor A3. The potentiometers R_{sense} were adjusted to cover the whole measurement range (depending on the ADC-resolution) for distance and force. The current for digital sensor D1 and D2 was set to 30 mA and to 200 mA for D3. As the current for all digital sensors D1 to D3 is pulsed (with pulse-widths of only a few hundred microseconds), the average current consumption was much smaller.

Distance and force measurements were taken from 1 adult male subject as follows. For **distance measurements**, spacers with lengths of $\{0, 2, 5, 10, 15, 20, 25, 30\}$ mm were attached to

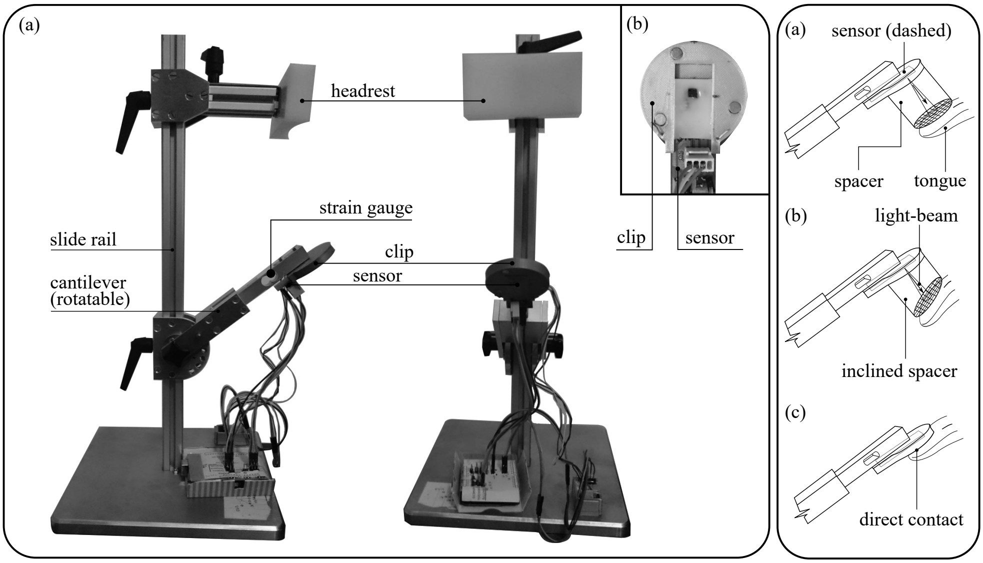

Figure 2 – Left: (a) Constructed test station. (b) Close-up of a sensor mounted on the sensor clip. Right: Spacer setup for the different measurement quantities. (a) Level spacer placed between sensor and tongue for distance measurements. (b) Inclined spacer for angle measurements. (c) Removed spacer for force measurements.

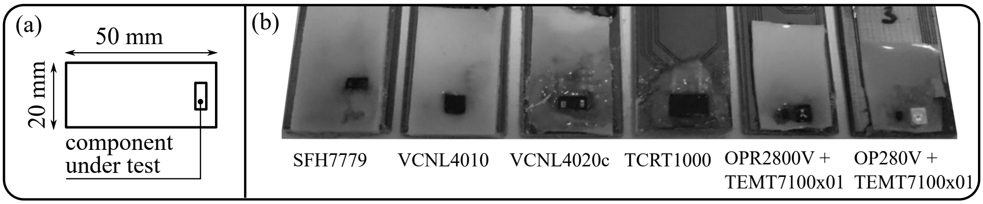

Figure 3 – (a) Standardized test sensor dimensions and optical component position, (b) examples of tested sensors and their encapsulation.

the sensor clip in sequence and the tongue was pressed very slightly against it to ensure contact (Figure 2 right, (a)). For each measurement, 200 samples were taken and the mean and standard deviation calculated. Inclination angle measurements were performed analogously, except the distance between the sensor and the tongue was fixed to 17 mm and 22 mm, respectively (as in [17]) and the inclination angle varied by $\{0, 5, 10, 15, 20, 25, 30, 35\}^\circ$, ($0^\circ$ corresponded to a light-beam orthogonal to the tongue's surface). For **force measurements**, the spacers were detached and the tongue was pressed directly against the sensor (Figure 2 right, (c)), slowly increasing the force from 0.1 N to 8 N during a time interval of 3 to 4 seconds and then decreasing the tongue force with the same, but opposite motion from 8 N to 0.1 N. 20 repetitions of increasing and decreasing the tongue force were performed for each sensor. The mean and standard deviation were calculated in a moving averager manner: all force- and ADC-measurements were sorted in ascending order and the mean and standard deviation was calculated in an interval of 10 values before incrementing the interval by one value.

5 Results and discussion

The ideal curve for both force and distance measurements would have a perfectly linear slope with zero variance in ADC-value for any given force or distance/angle value. Results for angle measurements are displayed in Figure 5. The actual performance over the whole dynamic range of distance and force measurements for all sensor combinations is shown in Figure 6. While the variance over all 200 samples for a single distance measurement (e.g., at 5 mm) was small across all sensors, force measurements varied quite significantly across consecutive readings (Figure 4, left). Additionally, a small hysteresis between the trajectories for rising and decreasing forces could be identified in most sensors as exemplary shown for the sensor D3 (Figure 4 right), where ADC-values were averaged as described in Section 4. Since the sensor performance for increasing force (and consequently pressure) measurements is much more relevant for logopedic exercises, only the values for *increasing* forces are considered.

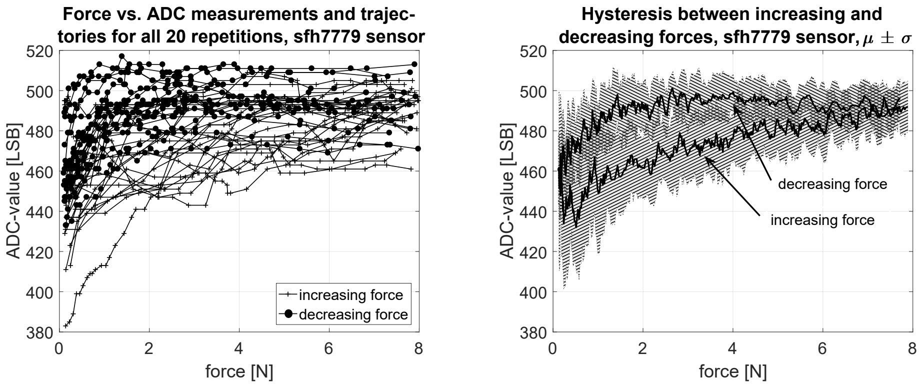

Figure 4 – Left: measurements of force vs. ADC-value across all 20 repetitions for sensor D3, separated for increasing (+) and decreasing (o) force. Right: mean and standard deviation (calculated as described in Section 4) for the emphasis of the hysteresis between increasing and decreasing forces.

Regarding distance measurement, the sensors performed quite as expected. Sensors with a narrow beam angle showed a steeper slope for measurements in the range of 20 mm to 30 mm, while the curves for all sensors with a wide beam angle declined rapidly towards the 10 mm mark. The measurements for distances > 15 mm for sensor D1 (d) and D2 (e) were still distinguishable from one another, however, this is difficult to see due to the huge range of ADC-values which span several magnitudes. It is important to mention that this fact was most likely due to the sophisticated timing and signal processing done directly on the chips. The analog sensor A3 which used a basic current-to-voltage circuit (see Figure 1) failed to resolve distances greater than 15 mm (as a result, sensor A3 was excluded from inclination angle measurements). The angle measurements were grouped into sensors with a wide half-power angle (around 60°) and a narrow half-power angle (< 25°). The results showed a slight overall increase in ADC-value with increasing inclination angle for sensors with a wide half-power angle, while the ADC-values for sensors with narrow half-power angles had the tendency to decline as inclination angle rose. These findings were consistent with [17]. Breathing against the sensor also had an influence on the ADC-value, because the sensor glass steamed-up, which became relevant at higher distances (> 20 mm) or lower absolute ADC-values, respectively.

Evaluating force measurements was more difficult, because of the high variance in ADC-values within - and across consecutive measurements. In terms of average slope linearity, sensor D1 (Figure 6, (d)) showed the best performance at the expense of the highest variance in ADC-value (with its ADC-resolution taken into account). Sensor A1, A3, D2 and D3 showed a somewhat similar behavior with a steep increase in ADC-value for low force-values and a gradual flattening towards higher force-values, while sensor A2 had an almost flat response to increasing force-values.

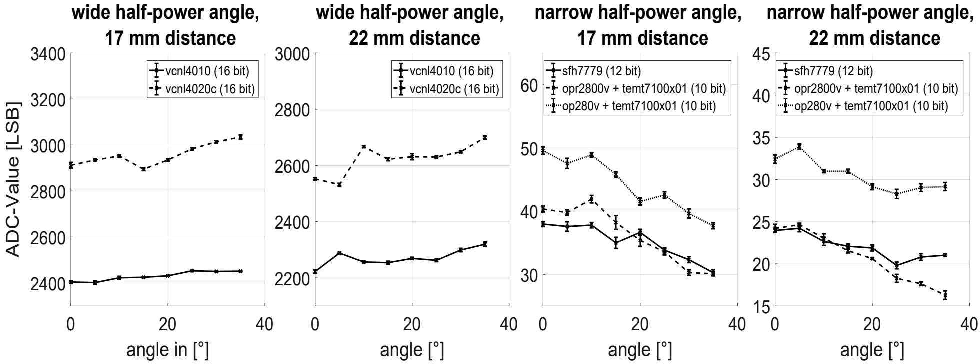

Figure 5 – Sensor ADC-value over inclination angle. The distance from sensor to tongue is fixed to 17 mm and 22 mm, respectively.

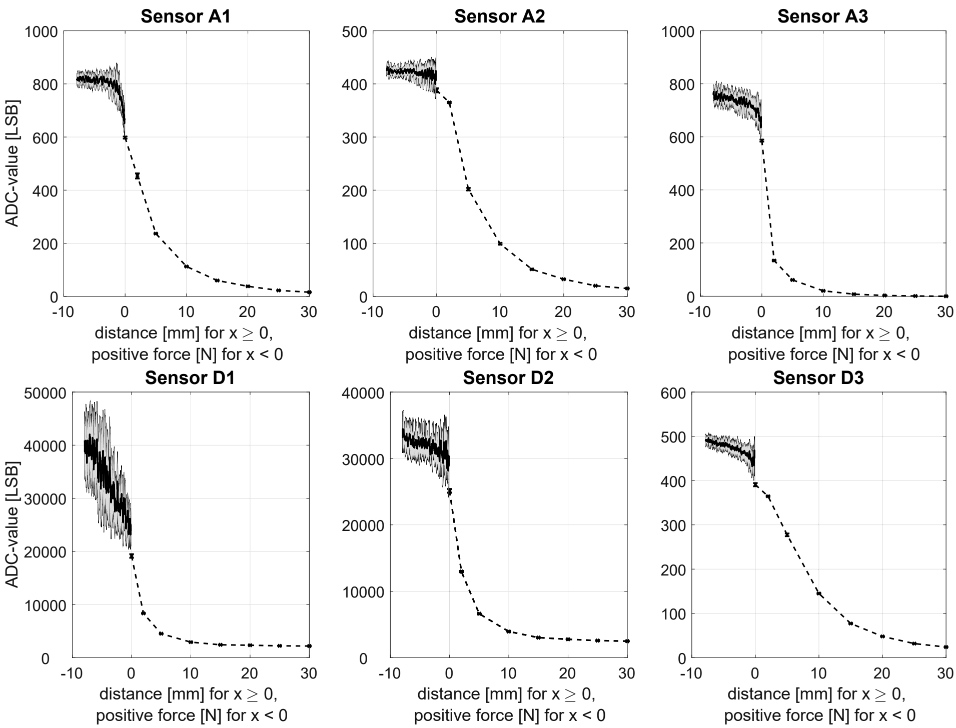

Figure 6 – Dynamic plots of combined force and distance measurements for all tested sensors.

Given these results, the actual *prediction* of whether a sensor is suitable for pressure sensing was equally difficult. Part of this is due to the fact that tongue tissue is quite complex and non-homogeneous and pressing different parts of the tongue against the same sensor already yielded different force measurement characteristics. Generally, however, light emitters with wide half-power angles showed slightly better performances in terms of slope linearity compared to the ones with narrow half-power angles, although sensor A1 and D3 have acceptable force-curves. The height difference between photo-detector and light source also seems to have an influence on force measurements and performance. Sensor D3 showed a much better force curve compared to sensor A1 and A2, both of which had their light source and photo-detector at different height levels, while all of the light sources had a narrow beam angle. Consequently, placing both components into one housing (as it is done in all digital sensors D1 to D3 and for A3) seemed to increase slope linearity and lead to a less erratic ADC-value trend as force increased. When evaluating the sensors with respect to their entire dynamic range (including force and distance), sensor D3 offered the best trade-off between a decent force characteristic and a good distance curve. Sensor D1 and possibly D2 could also be used, if the comparably worse distance characteristic could be improved with even further signal processing. Sensor A2 and A3 were unsuitable because of either their force (A2) - or distance (A3) characteristic and sensor A1 was composed of a laser-diode that is no longer in production and only included as a reference design.

6 Summary and Outlook

The developed process and test equipment served as an effective way to quickly evaluate optical components for their distance and force/pressure sensing capabilities. While the results for distance measurements were straight forward and as expected, the inherently complex tissue structure of the tongue made force measurements rather difficult and the governing parameters to predict sensor performance are still somewhat unknown. Nevertheless, the results show that pressure sensing is generally possible, although further insight into light scattering behaviour as well as appropriate signal processing and calibration methods are necessary to ensure the needed accuracy and precision. Further development of a device that will use optical distance and pressure sensing will thus focus on these topics.

7 Acknowledgements

This work has been funded by the German BMBF, support code 16SV7741.

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