Study of Static Tactile Detection Threshold via Pneumatically Driven Polydimethylsiloxane Membrane

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ABSTRACT
In this work we investigate how the tactile detection thresholds of air-driven actuators are influenced by design parameters as membrane thickness and diameter, when the finger is in contact with the deformable circular elastic membranes. We report lower thresholds with decreasing membrane thickness and increasing diameters. These results can help in the design of soft actuators for pin-arrays used as graphical tactile displays for visually impaired people.

General Terms
Tactile Displays, Pin arrays, Psychophysics, Pneumatics.

ACM Classification Keywords
H.5.2. Information interfaces and presentation: User Interfaces

INTRODUCTION
In general, the tactile display market was static in the recent years and no breakthrough is being observed. However, this topic attracts a lot of attention and many research centers and companies are dealing with large area tactile displays to make the digital world more accessible to the visually impaired community or to provide eyes-free computer-human interfaces. Information is generally organized as small pin-shaped actuators, named *taxels* as they can be considered the tactile equivalent of the pixel [1] Applications span from text reading to tactile graphics. Besides Braille bars, no portable device of reasonable cost exists, which have an area large enough to display graphical information, for example maps and scientific content.

In fact, commercially available systems use technologies which have very high costs per pin, due to the large number of individually assembled components. Cost linearly increases with the number of pins and grows exponentially when organizing pin in two-dimensional arrays.

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Soft actuation

Novel approaches [2,3] try to utilize advances in MEMS technology, to increase the power-to-weight ratio and create portable solutions. A growing interest is in the fabrication of *soft actuators*, i.e. actuators having lightweight mass, low damping coefficients and low stiffness [4]**.** They do have the big advantage of avoiding dot-by-dot assembly of the actuators, as a large area full of actuators can be produced at once using batch-processing technologies. Thus, the cost is, ideally, mainly defined by the number of required processing steps, regardless of the number of pins/actuators realized on the given surface.

Most of the MEMS-based soft actuation mechanisms have been applied to Braille geometries, mainly because Braille dots are small, but very few of the considered architectures are purely designed for tactile graphics, without necessarily providing text [5].

Beyond Braille

When dealing with perception of tactile graphics, as compared to Braille reading, the objective shifts from understanding information coded as single, separated small dots to picturing a sequence of symbols forming geometrical primitives. The specifications of Braille can therefore be relaxed: on the one hand, graphical raised line drawings can be perceived as low as 200 µm [6], while Braille requires dot elevation to be in the 600-900 µm range; on the other hand, larger dots are generally perceived better than small dots [7]. Specifically, large stroke seem not be a requirement when dots are larger than 2mm [8]. In line with that, response times in shape recognition using some three-dimensional tactile displays were inversely proportional to the object size. In addition, the third dimension was not reputed necessary compared to a purely bistable pin [5].

Therefore, the question is if one can build controllable actuators able to deliver sufficient force, enough stroke, without increasing too much the dot diameter, which would limit the resolution and as a consequence the number and type of displayed information.

Pneumatically-driven actuators [9, 10, 11] are ideal candidates to study such sufficient conditions, as they provide a wide range of low-to-high controllable forces.

Doh E., et al. in 2011 fabricated 3-axis tactile display actuator using PDMS pneumatic balloons for a robot assisted surgery system. The configuration could stimulate the finger with normal and shear forces, producing displacements of 1.5 mm in each direction [12].

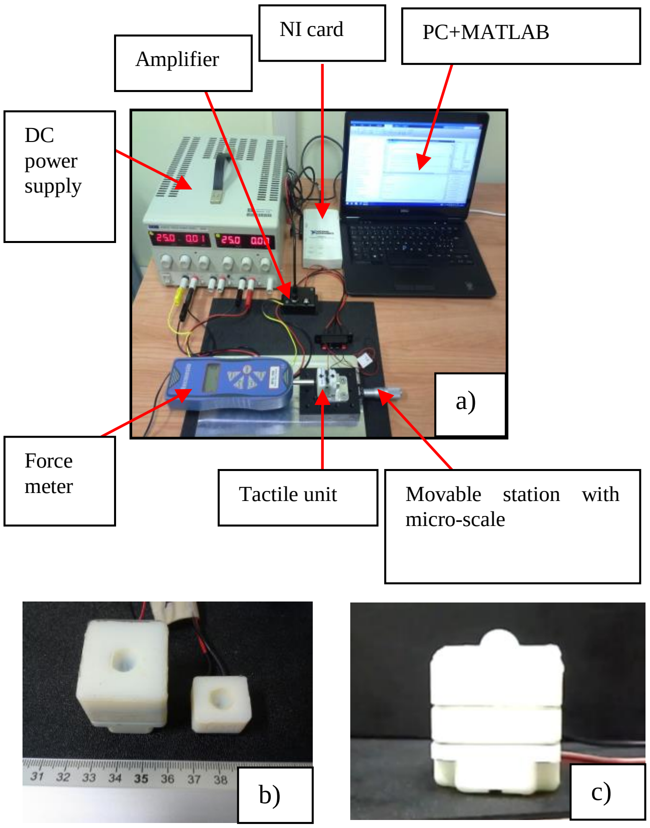

Figure 1. (a) Force-displacement setup (b) Single tactile unit with small and big cubes (c) Membrane inflated via micro-pump.

They can also very well mimic soft actuation systems, such as dielectric elastomers and shape memory alloys, for which the pressure below the actuator is constant, i.e. does not increase when being pressed by the finger. Constant pressure also allows to mimic *latching* properties i.e. the capability of the system to withstand the finger force with minimal energy consumption.

In addition, the kind and thickness of the membrane in contact with the finger is generally taken as a fixed parameter. Therefore it is generally unknown if varying thickness also varies perceived information.

In this preliminary study we investigate static tactile thresholds of single air-driven taxels, when the dot diameter is larger than Braille specifications, namely 4mm and 10mm. The first value corresponds to the inter-dot distance known to be optimal for roughness perception [8], the second is close to the two-point discrimination thresholds on the palm.

We also considered different thicknesses of the contact membrane, namely 200, 600, 1000 and 2000 µm. Our goal was to derive minimal physical requirements, in terms of force per area unit, so that a single dot is statistically distinguished from an otherwise smooth, non-rigid surface. We investigated possible dependencies from taxel diameter and membrane thickness.

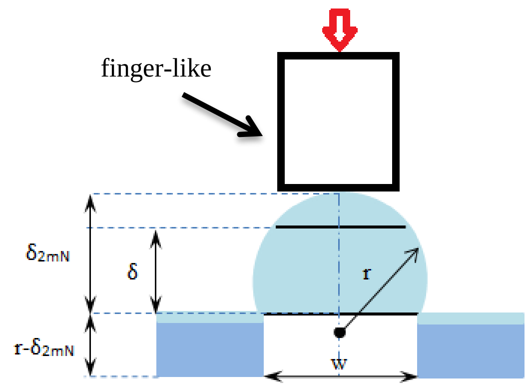

Figure 2. The model to convert displacements into contact areas.

EXPERIMENTAL SETUPS

The first experimental setup is aimed at measuring physical characteristics of taxels having variable diameter and contact membrane thickness. The second setup was aimed at estimating tactile detection thresholds when using the taxels of the first setup.

Materials

Ecoflex®50, Smooth-On, a Polydimethylsiloxane (PDMS) composite, was used as the membrane in contact with the finger. Elastosil 43, WACKER, an adhesive, was used to stick the PDMS membrane on top of the support, made of rapid prototyping material Verowhite 835. Rapid prototyping technology was used to fabricate taxels shaped as cuboids of variable size. Pneumatic energy was output from a piezolelectric micro-blower from Murata, able to reach pressures higher than 2kPa.

Tactile Unit Fabrication

Ecoflex®50 is two-component based PDMS, composed of A and B; the ratio of mixing (A:B) is 1:1. The mixture was spin-cast by 300 rpm for 30 s to create homogeneous surface on TEFLON petri-dish. The spin-cast mixture was treated in vacuum to avoid air bubble when it solidifies before curing overnight at 80 ºC. When membrane was ready, it was adhered on the Elastosil coated substrate, which was printed in two configuration: a first setup involved small cubes 19 x 21 x 12 mm , a second setup involved bigger cubes 30 x 33 x 32 mm (see Figure 1b and 1c). Then it was cured overnight at 50 ºC to obtain fully cured membrane and stable properties. The 3D-printed component, with the membrane on top, was wedged on the nozzle of the piezoelectric micro-blower. In the second setup (big cubes), a further 3D-printed component

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surrounded the micro-blower to minimize accidental motion of the nozzle.

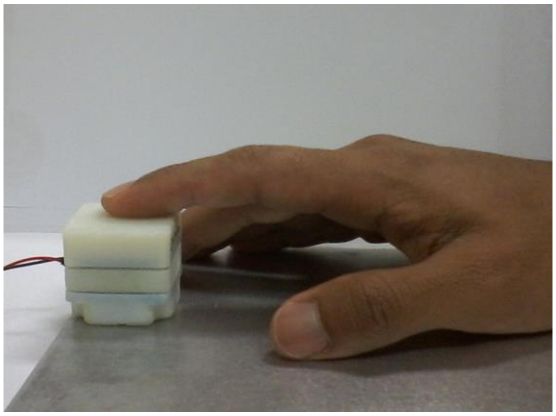

Figure 3. Psychophysical setup: the taxel fixed on a metal plate.

Force-displacement setup

The force-displacement setup is shown in Figure 1a. The tactile unit was inflated by the micro-pump. To produce the desired pressure, the micro-blower was voltage-driven at resonance by a sinusoidal wave, with fixed frequency 43 kHz and amplitude proportional to the desired pressure. We calibrated the voltage-pressure relationship to compensate for different micro-pumps, which exhibited repeteable but slighlty different characteristic functions. The micro-pump was supplied via DC power supply (TTi, Ex354 Tv, Triple Power Supply 300 W), and controlled by a MATLAB script (R2012b, 32 bit) through a National Instrument card (NI USB 6211, 16 inputs, 16 bit, 250 kS/s, Multifunction I/O), and a custom-built amplifier.

Preliminary measures were obtained with a DMA (Dynamic Mechanical Analyzer) to obtain force-displacement curves when force exceeded 1 mN (the DMA precision). Subsequent displacement measures were obtained with a moveable micro-scale station (THORLABS) and a force meter (Mecmesin, BFG 10 N). Specifically, a zero displacement is equivalent to a PDMS surface with 2 mN (the force-meter precision) of contact force with the force meter probe. The microblower was controlled to provide variable pressure values. For each pressure value, displacement was decreased up to zero with the micro-scale while measuring blocking forces.

Since no control on the actual pressure at the membrane was available (only the output from the micropump nozzle was known) we compensated possible air leakages out of the small/big cubes by normalizing pressure values fed to the psychophysical setup.

The true pressure values were estimated with the model depicted in Figure 2: we assumed the membrane deformation behaved as an inflating sphere of radius *r* given by:

$$r = \frac{\delta_{2mN}}{2} + \frac{w^2}{8\delta_{2mN}} \quad (1)$$

where r = radius of curvature; δ_{2mN} = displacement at 1or 2 mN (for the DMA and the force meter respectively) of preload wrt the reference surface; w = width of the hole (the taxel diameter).

When the finger or force meter exerted a force (red arrow) on the "on" taxel, the elasticity of PDMS allows the probe to flatten the sphere up to a distance δ from the background.

The flattened area is the probe-taxel contact area is given by:

$$\mathrm{A}(\delta) = [(r^2 - ((r - \delta_{max}) + \delta)^2] \quad (2)$$

and the estimated pressure is obtained by dividing the measured force by A(δ).

Psychophysical Setup

In a first psychophysical experiment, thought as a pilot study, we used the small cubes, to be held between the thumb and the middle finger. The index finger had to touch the PDMS membrane. In a second experiment (see Figure 3), participants had their dominant hand resting on an Aluminium plate, where the setup with big cubes was mounted and fixed.

Nineteen participants (thirteen women) had a mean age of 33 years (range 26-41). For each participant, 5 min practice preceded threshold estimation. Then, they underwent at least 6 tactile detection threshold estimation blocks to define for each participant their detection threshold (target level of 75% correct) using the method of constant stimuli. Each threshold estimation block comprised 30 trials in which tactile stimuli with different pressure levels were presented pseudorandomly either in a first or in a second interval lasting 3.5 seconds. Each pressure level was reached as described in the force-displacement setup.

First interval was defined by a single-beep sound. Second interval was signaled by a double-beep sound. Participants were instructed to touch the tactile unit with the index finger of their dominant hand each time they heard the sound and to verbally report whether they perceived a displacement of the membrane in the first or the second interval (two-interval forced choice detection task; 2IFC). Control condition was a zero displacement interval.

To estimate the threshold for 75% correct detection, we used the *psignifit* toolbox (psignifit.sourceforge.net) version 3.0 for Matlab, which implemented the maximum-likelihood method described in [13] for Weibull curve fitting.

When the estimated pressure of different setups resulted different, we normalized the thresholds for the relative efficiency. Unless otherwise stated all the reported statistics are one-tailed t-tests.

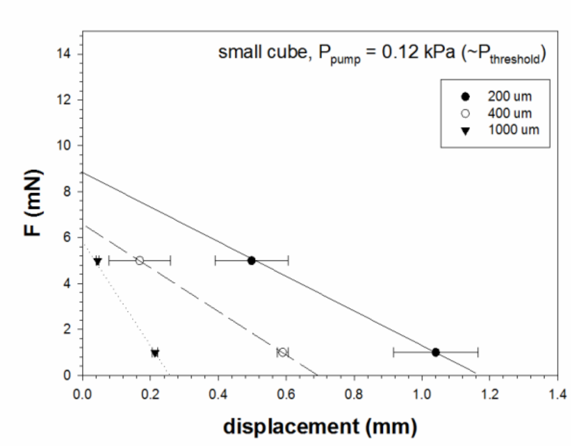

Figure 4 shows preliminary force-displacement curves with a taxel diameter of 10mm and variable thicknesses of the PDMS membrane.

RESULTS

Force-displacement Measures

As an example, Figure 4 shows preliminary force-displacement relationships, using a 10mm diameter taxel and variable membrane thicknesses. As expected, the blocking force increases as displacement decreases. Thinner membranes seem to transmit more force on equal displacements, or alternatively seem to be deflected more on equal forces. Qualitatively, thicker membranes also look stiffer. This is expected to have consequences at psychophysical level.

Figure 5(A), instead, shows the force-displacement curves obtained with two independent setups: a small and a big cube topped by a 200 µm membrane. The same was done with bigger cubes (not shown). The lowest forces (this time measured with the force meter) are limited to 2 mN. Figure 4 also shows that at the considered pressure (0.32 kPa), the displacement values are above known stroke thresholds (i.e. 200 µm for rigid mechanical couplings). The flattened-sphere model allows to derive Figure 5(B), which shows the force-area relationship.

Since a bias between the setups is apparent, the derivative of the force-area curves is taken between 0 and 10 mN to normalize subsequent psychophysical results. For example, the setup with the small cube, which appears more efficient, exhibits an overestimated pressure of a factor 1.72.

Psychophysical Results

We first compared tactile detection thresholds using small cubes with different membrane thicknesses. 75% correct detection levels (kPa) were entered into a repeated measures ANOVA with membrane thickness (200, 600, 1000 and 2000 µm) as within subjects factor. The ANOVA showed a main effect of thickness [$F(3,27) = 19.11$; $p =$

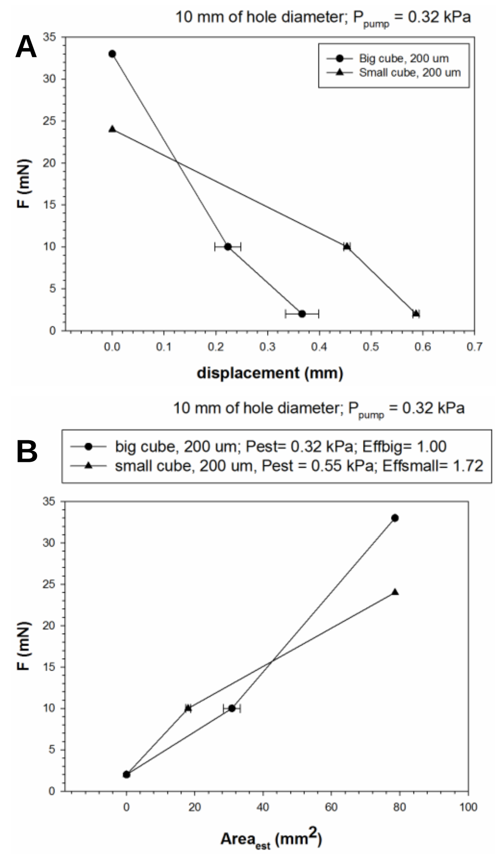

Figure 5. Force vs displacement (A) and equivalent force vs area (B) for two setups. The relative efficiency is estimated with the flattened sphere model.

.000001]. Follow-up one-tailed t-tests showed that detection thresholds significantly increase with membrane thickness (threshold for 200 µm: 0.05 kPa ± 0.01 SE; for 600 µm: 0.07 kPa ± 0.01; for 1000 µm: 0.1 kPa ± 0.01; for 2000 µm: 0.46 kPa ± 0.09; all p values < .03; see Figure 6).

Secondly, we analyzed results obtained with big cubes in which we manipulated target diameter and membrane thickness. We first compared tactile detection threshold performance when touching stimuli with different diameters after pooling over for membrane thicknesses. Threshold was significantly higher for 4 mm (5.36 kPa ± 0.85) compared to 10 mm target (0.10 kPa ± 0.01, $p < .0000001$). Then, we considered separately different membrane thicknesses. Detection thresholds are still significantly higher for 4 mm compared to 10 mm targets (for 200 µm thickness: 4.17 vs. 0.07 kPa; $p = .002$; for 1000 µm thickess: 6.16 vs. 0.12 kPa; $p = .00006$; see also Figure 7). On the contrary, detection threshold did not differ with different membrane thicknesses (threshold for 200 µm thickness: 1.89 kPa ± 0.86, threshold for 1000 µm

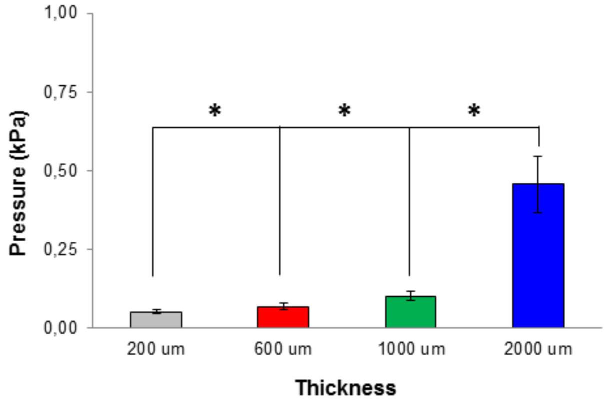

Figure 6. Tactile detection thresholds (75% correct detection level) for different membrane thicknesses using small cubes.

thickness: 2.90 kPa ± 1, Mann-Whitney U = 40.5; *p* = .23). However, when considering separately 10 mm and 4 mm diameter stimuli a different perceptual profile emerges. While the difference between 200 and 1000 µm thickness for the 4 mm diameter is still not significant (threshold for 200 µm thickness: 4.17 kPa ± 1.16, threshold for 1000 µm thickness: 6.16 kPa ± 1.14; *p* = .13; see Figure 7b), when considering 10 mm diameter stimuli, detection threshold is significantly lower for 200 µm compared to 1000 µm thickness (threshold for 200 µm: 0.07 kPa ± 0.008, threshold for 1000 µm: 0.12 kPa ± 0.01; *p* = .047; see Figure 7a).

DISCUSSION

In our first experiment we checked for possible influences of variable membrane thickness on tactile detection thresholds of a taxel with fixed diameter. Subjects detected more easily thin membranes. This means that mechanical energy is transferred more efficiently from an air chamber to the finger mechanoreceptors with as little transducing material as possible.

Force-displacement relationship indicates that part of the energy is dissipated in the elastic deformation of thicker membranes. Increasing thickness by a factor of 10 increased thresholds by a factor of 5. Thin membranes seem therefore to be preferable when building such actuation systems.

In our second experiment we checked for possible influences of taxel diameter on tactile detection thresholds, with varying thicknesses. When small-diameter taxels were "on" they were less distinguishable from their "off" state as opposed to large-diameter taxels. Reducing diameter by a factor of 2.5 increased thresholds by a factor of between 50 (using thicket membranes) and 60 (using thinner membranes).

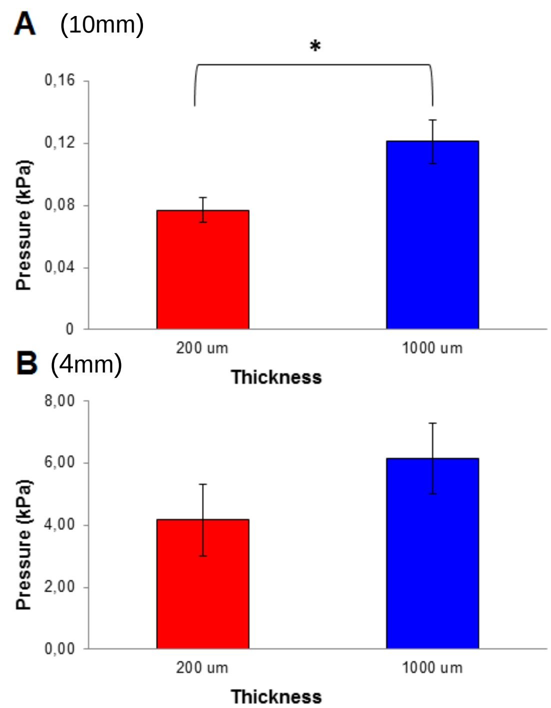

Figure 7. (A) Tactile detection thresholds (75% correct detection level) for different membrane thicknesses and 10 mm stimuli diameter using big cubes (B) Tactile detection thresholds for different membrane thicknesses and 4 mm stimuli diameter using big cubes.

A trend suggests that the perception of thicker membranes is more problematic with small diameters as well, even if more experiments are needed to confirm this aspect.

One could argue that to better understand what is the main factor underlying perceptual thresholds a constant diameter/thickness ratio should be considered (i.e. the larger taxel is a "zoomed" version of the smaller one). Yet, had we done so, to keep such ratio constant we would have considered thicknesses of 500um and 2000um for the larger taxels, for which, approximately, the perception thresholds of the first might have been three times higher (while with 4mm taxels it increases by 50% only). The higher sensitivity with larger taxels can be possibly explained by a combination of two other factors: a larger area of mechanoreceptors (mainly SA1) and a different finger/taxel mechanical coupling.

In fact, qualitative observations from our subjects report that the hole of the 3d-printed taxel (i.e. the contours of the free-standing membrane) was considered a cue to distinguish "on" from "off" taxels, meaning that the underlying structure and geometry of the rigid support can be very important.

The differences between our setups, which we attempted to compensate, may be due to air leakage, or to elastic properties of the Ecoflex membranes [14], which varied across time. Further data collections, averaging force-displacement curves across different taxels, will be necessary to clarify this aspect.

This study is important for designing perceivable taxels with technologies even beyond air-driven setups, i.e. whenever a taxel is powered without latching mechanisms and whenever pressure under a soft actuator can be assumed constant. Our results can also be useful for the design of *static* soft taxels, which have very different (generally more stringent) requirements as opposed to vibrating taxels. We emphasize that static taxels better approximate paper-based tactile maps, currently the standard for visually impaired subjects.

Overall, larger taxels with thinner membranes looked perceivable with less energy. However, this implies decreasing the resolution of any pin-array display. Trading off energetic requirements with resolution seems therefore a necessary step in the design of graphical pin-array tactile displays.

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