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A Tactile-Pattern-Integrated Sensing Window for More Consistent Photoplethysmography (PPG) Measurements

Changmok Choi, Jeongeun Hwang, Jongwook Lee, Byung-Hoon Ko, Youn-ho Kim, Hyuck Choo

Abstract—**We have demonstrated a tactile-pattern-integrated sensing window for more consistent photoplethysmogram (PPG) measurements. The pattern is composed of two tiny bumps that measure 500µm in diameter and 300µm in height and allow users to position their finger pulps more consistently on the sensing window over different measurement occasions, simply by following their tactile sensation. We experimentally compared the tactile pattern window to a flat window (without any bumps) for 5 test subjects and found that the sensing window with the tactile pattern significantly helped users obtain more consistent PPG signals than the flat window ($p < 0.01$).**

The use of PPG sensors in mobile phones and wearable watches have been limited to the measurements of heart rates and blood oxygen saturation in spite of widely-spread efforts to expand their applications. This is due to the fluctuations observed between measurements which largely originate from inconsistent placement of fingers on the sensing windows. The integrated tactile pattern could provide consistent and accurate measurements and lead to more successful commercialization of diverse PPG-based mobile healthcare services.

I. Introduction

Photoplethysmography (PPG) is a noninvasive optical measurement which detects blood volume changes. It has long been used in hospitals to detect heart rates and blood oxygen saturation (SpO_2) [1]. PPG could also provide other important vital parameters. In recent years, more diverse PPG-based applications using mobile devices have been explored with promising results: smartphone-based measurements of left ventricular ejection fraction from the right carotid artery [2]; estimation of the sleep quality using a wearable pulse oximeter [3]; smartphone-based respiratory rate measurements in conjunction with ECG sensor [4]; quantitative algorithmic assessment of vascular health from the fingertip [5]. However, despite these efforts, the main use of PPG in mobile devices has been limited to the measurements of heart rates and SpO_2 for wellness purposes.

For SpO_2 measurements in a hospital, clip-on PPG devices are used, and their upper and lower clips press on the upper and lower sides, respectively, of a finger. Two LEDs of 660-nm and 940-nm wavelengths are located inside the upper clip while a photodiode (PD) is located inside the lower clip. The light from two LEDs penetrates finger tissues, passes through arteries of the finger, and arrives at the PD. The upper and lower clips are linked together with a spring to apply an appropriate and constant level of pressure on the finger for consistent measurements.

C. Choi is with Samsung Electronics Co., Ltd., 129 Samsung-ro, Yeongtong-gu, Suwon-si, Gyeonggi-do, 16677, Republic of Korea (e-mail: cm7.choi@samsung.com).

J. Hwang, J. Lee, B. Ko, Y. Kim, and H. Choo are with Samsung Electronics Co., Ltd., 129 Samsung-ro, Yeongtong-gu, Suwon-si, Gyeonggi-do, 16677, Republic of Korea.

In contrast, PPG sensors in mobile devices have LEDs and PDs on the same side, so the light intensity measured by the PD represents the amount of reflection by the blood in the finger. Because of this placement of the LEDs and PD, the measured light is heavily influenced by the degree of the tissue deformation which consequently determines how deep the photons would travel into the finger: the epidermis, reticular dermis, or hypodermis [6].

The penetration depth of photons significantly influences the accuracy of a SpO_2 measurement because the SpO_2 level is defined as an estimation of the functional hemoglobin fraction in the artery that is saturated with oxygen. This measurement is made by a pulse oximeter. If the users are not properly guided to position their finger pulps on the sensor (Fig. 1), the position of their fingers will be different for each measurement. The directional mechanical stiffness of the finger tissue as well as the applied contact force between the finger pulp and the sensor surface will be different depending on the contact position. In addition, different contact forces will lead to different deformation of the tissue, so the penetration and travelling path of light in the finger will change, and this will lead to inconsistent, fluctuating measurements even for the same user in a stable physical condition.

For successful commercialization of diverse PPG-based mobile healthcare services, it is very important to make a deterministic model by using an input of a PPG feature vector to estimate an output as a diagnosed value or status. In the deterministic model, no randomness is involved in the development of future states of the model, therefore, the deterministic model will always produce the same diagnosed values or statuses from the same PPG feature vectors. A PPG sensor of the mobile device currently available in the market lacks the ability to repeatedly obtain the same PPG feature vectors as the user's diagnosed medical status. PPG waveforms would vary depending on the measurement position of a hand [7]. Therefore, the contact position of the skin must be well-guided for the user.

Being inspired by braille, a tactile writing system, we integrated a tactile pattern on the sensing window as a physical placement guide for users. When human touches an object by a hand, the delicate tactile perception of the finger pulp can differentiate and remember the spatial texture information of object surfaces [8]. We have demonstrated experimentally that the PPG sensing window with two tiny bumps guided the users to position their finger pulps more precisely and consistently on the center of the PPG sensor, showing superior effectiveness of the integrated tactile pattern over that of a flat surface without bumps. The integrated tactile pattern approach promises more consistent measurements for diverse mobile PPG applications and may open up a shortcut to commercialization.

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II. Materials and Methods

A. Design of the PPG sensor cover

In Fig. 1, the PPG sensor consists of a sensor block to protect a PPG sensor and a printed circuit board, pieces of window where light can transmit, and two bumps that measure 500μm in diameter and 300μm in height to provide tactile stimuli to the finger pulp.

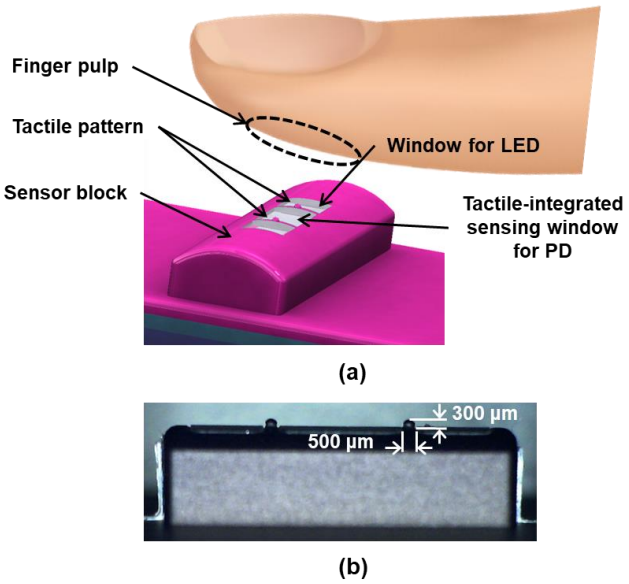

Figure 1. (a) Schematic design of the developed sensing window. (b) The size of the tactile pattern.

B. Experiment design

Since the magnitude of the contact force between the finger pulp and the PPG's sensing window will influence the shape of the PPG waveforms [9], the same contact force was induced to the finger pulp of all subjects using a 7 degrees-of-freedom robot device (Omega 7 by Force dimension) as shown in Fig. 2. A 6 DOF force/torque sensor (NANO 17 by ATI Industrial Automation) was installed below the sensor interface, and the measured force was fed back to the robot device to accurately control the contact force.

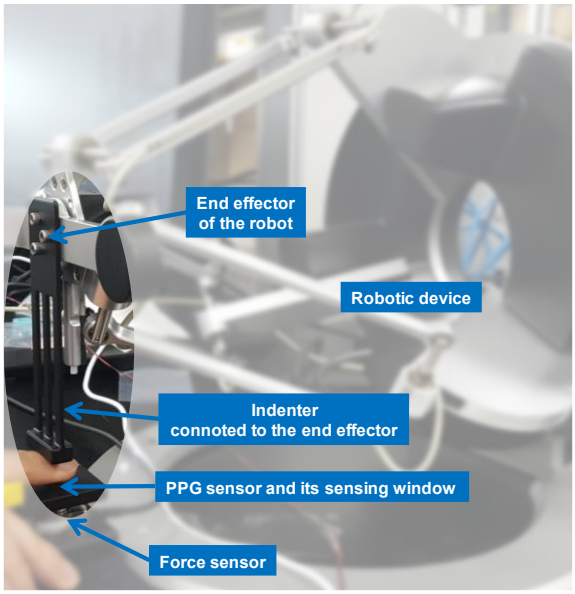

Figure 2. Experimental setup.

C. Experimental protocol

Five test subjects (mean age of 35.6 years) volunteered for this study. All subjects had no known cardiovascular, neurological, or respiratory diseases, and they were fully informed of the experimental procedure. All subjects were asked to comfortably sit on a chair, and place their index finger pulp on the center of the sensing window. Figure 3(a) shows a flow chart of the experimental protocol. Firstly, a flat window without bumps was used to measure PPG. The robot induced and held 1-N contact force on the finger during the measurement that lasted one minute (Trial 1). After the first trial, the subject was asked to take his/her finger completely off the sensing window, and place the finger pulp on the center of the sensing window again. This PPG measurement was repeated twice more (Trial 2 – Trial 3) using the same flat window. After three trials, we used the tactile-pattern-

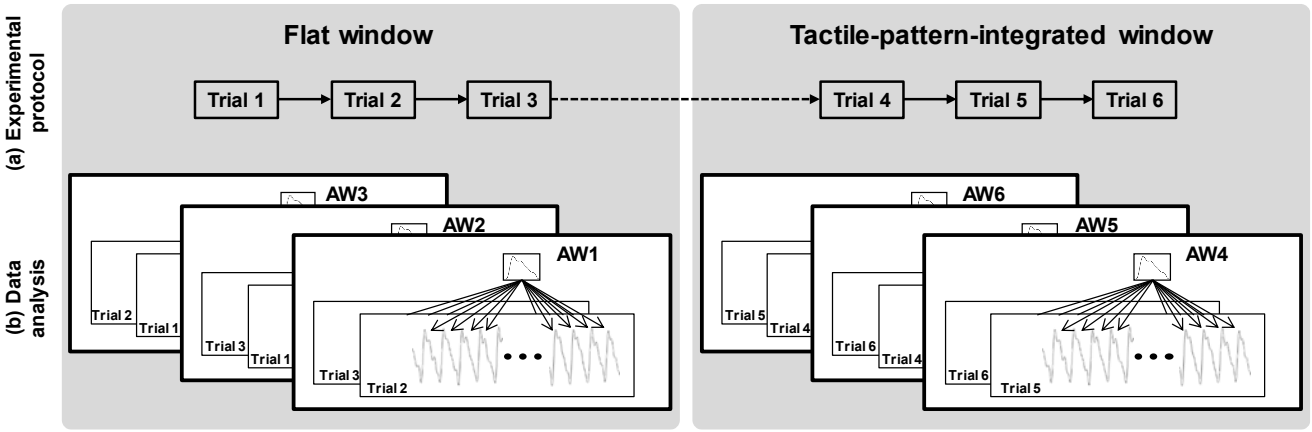

Figure 3. A schematic representation for the (a) expermental protocol and (b) data analysis for one subject.

integrated window on the same subject following the same protocol. We designed this protocol to compare the consistency of the PPG waveforms between Trial 1 – Trial 3 for the flat window and Trial 4 – Trial 6 for the tactile-integrated window.

D. Data analysis

We used a high pass filter on the PPG data to remove a baseline wander caused by the respiration of the test subjects and low pass filter to remove high frequency noises. Pulse-wave peaks were extracted using the method described in Reference 10, and each pulse waveform was normalized in time and magnitude.

E. Statistical analysis

An averaged waveform (AW) of PPG was extracted from each trial by averaging 20 pulse waveforms in a trial. Each AW was compared to 40 waveforms of the different trials in the same subject using the same window as shown in Fig. 3 (b). Correlation coefficient (CORR) and root-mean-squared error (RMSE) were obtained between the AW of the i-th trial and the individual waveforms of j-th trial ($i \neq j$) for the flat window (Trial 1 – Trial 3) and for the tactile-pattern-integrated window (Trial 4 – Trial 6), respectively. The significance of differences in the obtained CORRs and RMSEs between two windows were analyzed by Wilcoxon rank-sum test. $P < 0.01$ was considered statistically significant.

III. Results

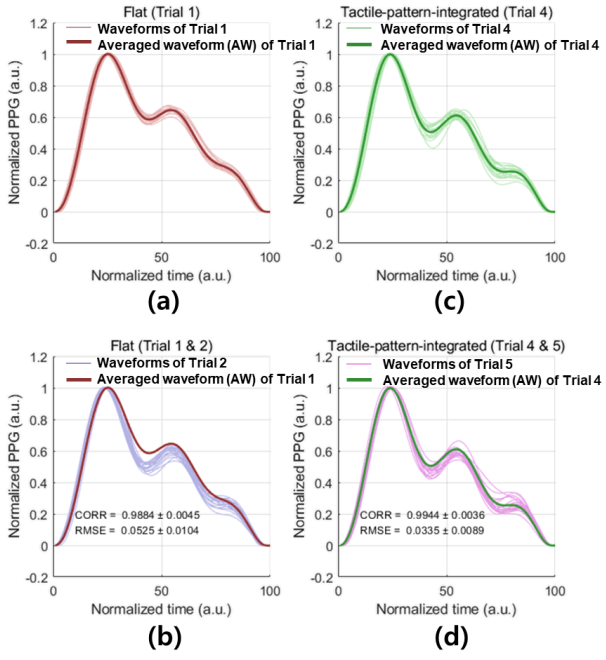

Figure 4. An example of measured PPG pulsewaves from one subject: (a) 20 waveforms of Trial 1 (thin red) & the AW of Trial 1 (bold red), (b) 20 waveforms of Trial 2 (thin blue) & the AW of Trial 1 (bold red), (c) 20 waveforms of Trial 4 (thin green) & the AW of Trial 4 (bold green), (d) 20 waveforms of Trial 5 (thin pink) & the AW of Trial 4 (bold green).

Figure 4 shows an example of the effectiveness for the tactile-pattern integrated window against the flat window. In Fig. 4 (a) and (c), 20 PPG waveforms were plotted for the first trial and their AW using the flat window and the tactile-pattern-integrated window, respectively. After the first trial, the subject took his/her finger completely off the sensing window, and then he/she place the finger again on the center of the window for the next trial. If the subject place the finger on the near position in different trials, PPG waveforms in different trials should look similar.

For a case of the flat window, Fig. 4 (b) shows CORR = 0.9884 ±0.0045 & RMSE = 0.0525 ±0.0104 between two different trials. For a case of the tactile-pattern-integrated window, Fig. 4 (d) shows CORR = 0.9944 ±0.0036 & RMSE = 0.0335 ±0.0089 between two different trials. Since the tactile-pattern-integrated window have the greater CORR and less RMSE, the PPG waveforms are clearly more consistent between the two trials than the flat window.

We evaluated CORRs and RMSEs for all the subjects. The average CORRs were 0.9917 ± 0.0059 and 0.9938 ± 0.0046 for the flat window and tactile-pattern-integrated window, respectively. The average RMSEs were 0.0474 ± 0.0175 and 0.0376 ± 0.0150 for the flat window and tactile-pattern-integrated window, respectively. Figure 5 shows a box plot with the comparison results of CORRs and RMSEs between the flat window and the tactile-pattern-integrated window. The CORR of the tactile-pattern-integrated window was significantly higher than that of the flat window (p = 4.59E-12). The RMSE of the tactile-pattern-integrated window was significantly lower than that of the flat window (p = 8.93E-25). Therefore, we can conclude that the tactile-pattern-integrated window helped obtain more consistent PPG waveforms from the subjects in the stable

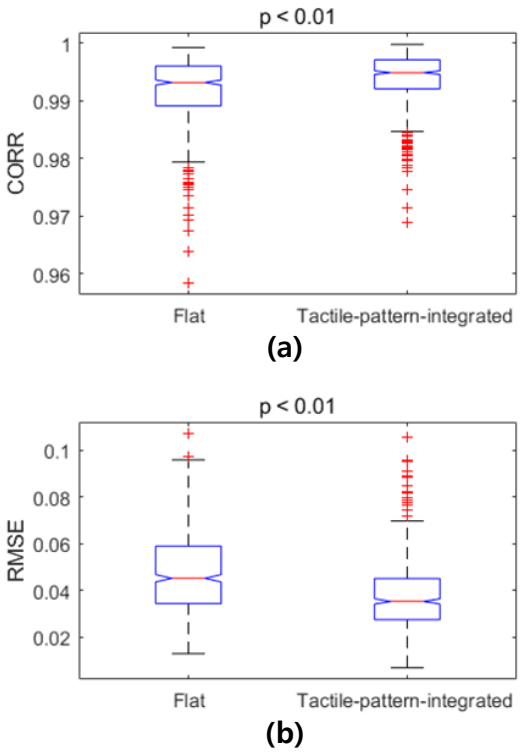

Figure 5. Box plots to compare the PPG-waveform similarities among different measurement occassions for the flat window and the tactile-pattern-integrated window: (a) CORR and (b) RMSE.

health condition at different measurement occasions.

IV. Concluding Remarks

Today, smart phones are the most indispensable gadget to modern people. Its main function is of course wireless communication, but it opens a door to implement a great mobile healthcare platform. First, most people have at least one smartphone, and they bring it with them everywhere. Therefore, if the smart phone has appropriate health functions, people could check their health anywhere and at any time. Second, computing power of smart phones have been improving fast. A microprocessor of the phone can provide high speed computation with low power consumption and determine health status from the data measured using the phone. In addition, the large size of the onboard smartphone memory can store the past history of the biomedical data and analysis results. Third, the smart phone can connect to the cellular network almost anywhere and at any time, so the network service can provide an important link between the user and the appropriate medical experts. Moreover, a GPS sensor of the phone can provide a location of the user in an emergency situation to the paramedics. Lastly, a variety of small sensors are already installed in the phone such as an accelerometer, PPG, a microphone, and cameras, and more sensors are expected to be included in the near future. The new sensors will measure a wide range of biomedical signals from the human body for quantitative and objective medical diagnoses.

PPG sensors in current smartphones can utilize multiple wavelengths of light and expand their applications, but mostly available services provide the measurements of heart rates and SpO_2 levels. There is a worldwide race to develop a new mobile healthcare application to provide medical services based on PPG. The success of the newly developed application will require a deterministic model that maps the input of a PPG feature vector to the output as a diagnosed value or status. For the development of the deterministic model, there are currently two challenges.

One challenge is the lack of opto-physiological knowledge involving PPG. Most researchers currently believe PPG signals represent only volumetric values to the changes of the blood vessels in the tissues. However, the PPG signals are not only dependent on the amount of red blood cells but also mechanical properties of the blood vessels and soft tissues, wavelengths of PPG, and distances between a light source and a detector [11] [12]. The opto-physiological knowledge of PPG should be more carefully studied to develop the deterministic model.

The other challenge is the large fluctuation between measurements even for the same test subject in the stable physical condition, resulting in noisy and sometimes irreproducible PPG measurements. Kamshilin *et al.* reported that PPG waveforms are different depending on the measurement position of the hand [7]. Measurement fluctuations will add randomness to the input data and confuse the pre-defined model. Such randomness will interfere not only with estimating accurate health status of the user from PPG readouts but also prevent from developing a deterministic model based on the large amount of the measured data even from a single user. If a clip-type interface is used to measure PPG signals in a mobile device, randomness can be reduced because the measurement position can be well-guided. However, mobile devices are always forced to be designed as compact as possible with an exception of the display. Therefore, a solution to obtain stable and consistent PPG signals must be achieved without changing the general design of the mobile devices in the market.

In our study, the integrated tactile pattern does not alter the design of the conventional mobile devices. Thanks to the acute and surprisingly consistent tactile sensation of the human finger pulp, tiny two bumps on the PPG-sensing window enable the subjects to consistently position their finger pulp on the same position of the sensor cover at different measurement occasions and significantly improve the measurement consistency.

References

[1] J. Allen, "Photoplethysmography and its application in clinical physiological measurement," *Physiological measurement*. vol.28, no.3, R1, 2007.

[2] N. M. Pahlevan, *et al.*, "Noninvasive iphone measurement of left ventricular ejection fraction using intrinsic frequency methodology," *Critical care medicine*, vol.45, no.7, pp.1115-1120, 2017.

[3] S. Cheng and J. Huang, "An intelligent sleep quality detection system based on wearable device," *International Conference on Machine Learning and Cybernetics (ICMLC)*, 2017, pp.493-498.

[4] D. A. Birrenkott, M. A. F. Pimentel, P. J. Watkinson and D. A. Clifton, "A Robust Fusion Model for Estimating Respiratory Rate From Photoplethysmography and Electrocardiography," *IEEE Transactions on Biomedical Engineering*, vol. 65, no. 9, pp. 2033-2041, 2018.

[5] H. Wu, B. Lin, C. Yang, Y. Ou and C. Sun, "Assessment of Vascular Health With Photoplethysmographic Waveforms From the Fingertip," in *IEEE Journal of Biomedical and Health Informatics*, vol. 21, no. 2, pp. 382-386, 2017.

[6] A.V. Moço, S. Stuijk, and G. Haan, "New insights into the origin of remote PPG signals in visible light and infrared," *Scientific reports*, vol.8, no.8501, pp.1-15, 2018.

[7] A. A. Kamshilin *et al.* "A new look at the essence of the imaging photoplethysmography," *Scientific reports*, vol. 5, no. 10494, pp.1-9, 2015.

[8] R. S. Johansson and J. R. Flanagan, "Coding and use of tactile signals from the fingertips in object manipulation tasks," *Nature Reviews Neuroscience*, vol. 10, no. 5, pp.345-359, 2009.

[9] A. Chandrasekhar *et al.*, "Smartphone-based blood pressure monitoring via the oscillometric finger-pressing method," *Science translational medicine*, vol. 10, eaap8674, 2018.

[10] C. Choi *et al.*, "PPG pulse direction determination algorithm for PPG waveform inversion by wrist rotation," *39th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC),* pp. 4090-4093, 2017.

[11] H. Njoum and P. A. Kyriacou, "In vitro validation of measurement of volume elastic modulus using photoplethysmography," *Medical engineering & physics*, vol. 52, pp. 10-21, 2018.

[12] H. Njoum and P. A. Kyriacou, "Photoplethysmography for the Assessment of Haemorheology," *Scientific reports*, vol. 7, no. 1, pp. 1406, 2017.