A Combined Photoplethysmography and Force Sensor Prototype for Improved Pulse Waveform Analysis

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***Abstract*—Pulse detection has in recent years received extensive attention in physiological health monitoring, and continuous blood pressure estimation models based on pulse waveform analysis have shown great promises. Two widely used technologies for noninvasive detection of the pulse waveform are photoplethysmography and tonometry. Although the signals from PPG and tonometry are rooted in different physiological mechanisms, the technologies are typically used separately. The simplistic approach of estimating cardiovascular responses based on single sensor sources is discorrelated with the complexity of the human body. To acknowledge this, we have developed a low-cost, low-power sensor prototype combining PPG and force-sensing capabilities. The proposed solution is able to accurately detect the pulse waveform, both optically and tactually, in addition to estimating the sensor interface pressure. We hypothesize that the combined solution can be used to ameliorate one or more of the challenges associated with the separate technologies in physiological monitoring, and thus potentially give continuous pulse waveform based blood pressure estimation models valuable additional inputs, ultimately leading to increased estimation accuracies and measurement confidence.**

I. Introduction

In 2016, cardiovascular diseases (CVDs) represented 31% of all deaths worldwide [1]. High blood pressure is one of the main contributors to the burden of CVDs. Despite the devastating statistic, it is estimated that most premature deaths related to blood pressure and CVDs are preventable [2]. This makes noninvasive and continuous blood pressure devices a major target for improved health and wellbeing worldwide. In recent years, pulse monitoring by photoplethysmography (PPG) and force sensing has received extensive attention in physiological health monitoring because of their convenience and ease-of-use.

PPG obtains pulse information by optically detecting blood volume changes in the microvascular bed of tissue [3]. Although the small variations of light intensity are associated with blood volume changes, it has been demonstrated that the response contains similar information to that of the peripheral arterial pressure pulse [4]. Analysis of the pulse waveform obtained from PPG measurements has therefore emerged as a promising method to analyze cardiovascular health and estimate blood pressure [5], [6], [7]. One of the most valuable parameters in the analysis of the PPG contour is the PPG amplitude (PPGA), shown superior to other features for continuous blood pressure estimation [8]. However, the complex interaction of light with biological tissue poses a major challenge with the use of PPGA, and there does not exist any internationally recognized standards for the use of PPG, nor any calibration procedure to standardize PPGA between different measurements [9], [10]. Therefore, it is often difficult to know whether PPGA is affected by blood volume changes or any other factors, such as sensor interface pressure, amount of incident light, the orientation of red blood cells, or blood vessel wall movement [9], [10].

Force sensing, or tonometry, on the other hand, translates the slight force changes from the arterial pulsations into electrical signals. Compared to the not fully understood PPG technology, the signal from tonometers are easier to grasp as it is rooted in physical forces [9]. Despite the possibility of acquiring intra-arterial pressures directly from force recordings, the use of the technology is possibly limited by sensor elements having low sensitivity, high cost, and high energy consumption, in addition to being highly sensitive to positioning [11], [12]. Furthermore, the attenuation and distortion of the arterial pressure wave through the tissue can be complex, especially since both tissue composition and distance between the location of measurement and the intra-arterial site can be non-stationary and subject dependent.

Today, PPG and tonometry are typically used separately for pulse detection and pulse waveform analysis. The simplistic approach of estimating cardiovascular responses based on single sensor sources is discorrelated with the complexity of the human body, especially when considering the challenges with each of the technologies. In this article, we, therefore, suggest combining PPG and force detecting sensors. We hypothesize that the combination of the technologies can be used to ameliorate one or more of the challenges presented above. For example, a force sensor array can be used to obtain the PPG interface pressure. Furthermore, since the pressure from the artery is inversely related to blood volume, force readings can possibly be used as an additional input to determine if PPGA is affected by blood volume changes or any other factors [13]. Or, in the opposite way, PPG recordings can be used to cross-validate detected force changes. Therefore, we believe that the combined sensor prototype has the potential of providing pulse waveform-based blood pressure estimation models valuable additional inputs. This may lead to increased estimation accuracy and higher confidence in the detection of physiological responses [14].

II. Sensing Methodology

A. Implementation of Sensor Solution

The proposed sensor array, shown in Fig. 1, consists of four piezoresistive barometric pressure sensors (BMP388, Bosch Sensortec) and one PPG sensor (MAX30102, Maxim Integrated) mounted on a custom printed circuit board (PCB) of size 20 mm by 15 mm. The original purpose of a barometric pressure sensor is to measure atmospheric pressure changes. Its application can, however, be repurposed to detect forces by embedding the sensor elements in an elastomer [15]. By doing so, the elastomer forms a robust and compliant surface for contact and serves to transfer pressure changes within the layers of the elastomer to the pressure transducer. Rubber-embedded barometric pressure sensors have previously demonstrated excellent linearity, fast response, low noise, and negligible hysteresis [16].

In our work, the pressure sensor elements are embedded in a soft silicone rubber (Ecoflex 00-10, Smooth-On) with a thickness of 1.55 mm, the same as the height of the PPG element. The soft silicone rubber was chosen because attenuation and distortion of pressure waves are typically reduced in softer materials [17]. The surface layer of the sensor is covered by the same silicone rubber mixed with graphite powder (Fig. 1b), a material known to absorb light well [18]. This is done in order to reduce the interference of incident light to the PPG sensor. To increase the pressure sensor sensitivity, the diaphragm of each individual pressure sensor is opened to expose all off the pressure transducer element to the elastomer.

The complete sensor prototype costs roughly \$20 to build in individual quantities. The most expensive parts are the PPG sensor (~\$9) and the pressure elements (~\$4 each). Undoubtedly, a mass-produced version of the sensor could be made at a lower cost. The power consumption of the sensor solution was measured when worn and fully functioning. The result indicated a power consumption of 46.5 mW when the LED on the PPG sensor was set to maximum brightness and 22 mW when it was turned off. For reference, the largest Apple Watch Series 4 contains a 1.14 Wh battery. This suggests that the sensor can be integrated into future wearable electronics.

B. Force Sensor Calibration

The barometric pressure sensor array needs to be calibrated in order to map the pressure signals to represent force data properly. Since the sensor solution is to be compressed to the skin of the user, it is assumed that the forces of interest mainly will act normal to the sensing elements. Therefore, the pressure elements are calibrated to detect forces in the normal direction. Characterization data for calibration was acquired by compressing the sensor array onto a reference force sensor (FSAGPNXX1.5LCAC5, Honeywell) for ground-truth values.

Since the detection of the pulsating forces from the arteries are highly local, each of the pressure elements was calibrated separately. A number of different models mapping the sensor readings to the reference was tested through the System Identification Toolbox in MATLAB™. For the different sensor elements, the best result was obtained by individual 4th order discrete-time transfer functions. The calibrated sensor readings were validated by comparing the calibrated values to reference force data for compressions mimicking pulsations from the arteries, yielding a root mean square error (RMSE) for the different sensors of 0.0125N, 0.0071N, 0.0171N, and 0.0086N. The estimated forces are plotted in Fig. 2.

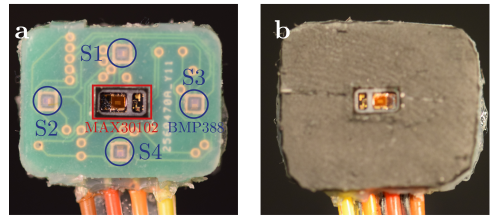

Fig. 1. **a**, Top view of sensor prototype (without graphite layer) with components highlighted. S1, S2, S3, and S4 indicates the position of PPT sensor element 1, 2, 3, and 4, respectively. **b**, Top view of sensor prototype with the graphite-induced silicone layer.

III. Physiological Sensing

As a proof-of-concept, the sensor's ability to obtain physiological information was tested by placing the sensor at the skin above the radial artery of a healthy 29-year-old male subject. The LED of the PPG sensor was set to illuminate the skin using infrared light (890 nm) with the PPG's internal photodetector detecting its reflections. A sample of the proof-of-concept recording is plotted in Fig. 3. Here, the PPG signal was filtered with a five-level orthonormal Meyer discrete wavelet filter, removing frequencies below 0.38 Hz. The baseline wander of the signal was further removed by fitting a cubic spline to the pulse onsets and subtracting it from the signal. An unequal baseline is seen as a typical reason for misextraction of PPG derived features [19]. In Fig. 4, the mean force of each cardiac cycle from S3 is plotted against PPGA, illustrating a slight inverse trend between the measures. S3 was chosen because it shows all the typical characteristics of the pressure waveform [20]. In Fig. 5, an interface pressure estimate is indicated, where the represented forces are linearly interpolated using the average of the force recordings in Fig. 3.

IV. Discussion and Future Works

The proof-of-concept results show that the combined PPG and force sensor are capable of obtaining information that potentially can be used to ameliorate one or more of the challenges associated with the separate technologies: (i) the force sensors enable estimation of a sensor interface pressure; (ii) PPG recordings capture pulse volume changes accurately; and (iii) the force sensing elements are sensitive enough to detect arterial pressure changes, enabling cross-validation of PPGA and force responses.

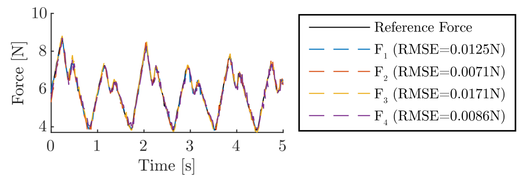

Fig. 2. Comparison of calibrated sensor forces with the reference data. The graph shows the calibrated force for each of the sensor elements plotted together with the reference recording. F_1, F_2, F_3, and F_4 relates to pressure sensor element S1, S2, S3, and S4, respectively, as represented in Fig. 1. RMSE for each of the calibrated forces are given in the description box.

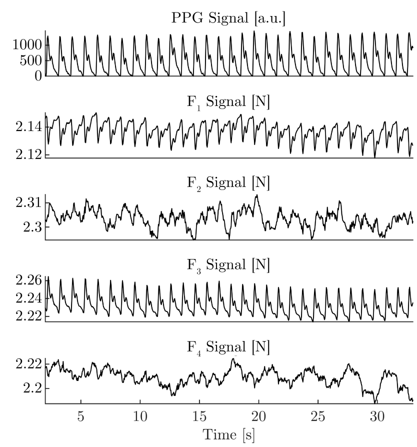

Fig. 3. Detected signal from proof-of-concept test. F_1, F_2, F_3, and F_4 are the respective forces detected by S1, S2, S3, and S4 as given in Fig. 1. From the graphs, F_3 illustrates a force waveform with all the typical characteristics.

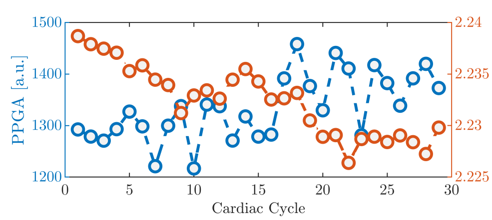

Fig. 4. PPGA compared with the mean force of each cardiac cycle of F_3 as presented in Fig. 2.

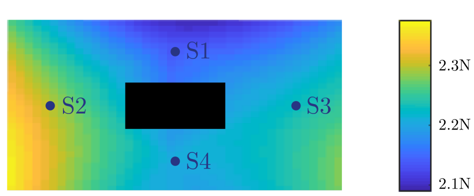

Fig. 5. Sensor interface pressure. The force distribution is estimated by linear interpolation of the mean forces of each of the force signals presented in Fig. 3.

With that being said, it should be noted that although a slight inverse trend between PPGA and the mean force of S3 is observed, the inverse relation is violated on a beat-to-beat basis almost as often as it is followed. It is unclear whether this is caused by small PPG perturbations (e.g. vessel wall movements, incident light, or the orientation of red blood cells) or if it represents actual blood volume changes. At the same time, it seems evident that the responses from the technologies, in fact, represent different physiological mechanisms and that they can be used advantageously in correlation. The potential benefits of combining the technologies should, however, be quantified and studied further, and tests relating the combined sensor readings to high-fidelity blood pressure measurements are underway.

The initial idea of the configuration of the pressure elements was to obtain pulse information from different locations, enabling triangulation of local recordings to obtain absolute pressure, in addition to estimating the sensor interface pressure. A serendipitous finding from the response in Fig. 3 was the response of S1 (F_1), showing an inverse response to that of S3. This can potentially be explained by the site-specific compressions by the arterial pulse, causing strain on areas in close proximity as a result of the conservation of mass of the silicone. This suggests that highly local information of the pulse may be obtainable with the proposed sensor element. Having said that, only S1 and S3 obtained a signal resembling the pulse pressure contour, so the configuration was experienced to be too widespread for the use of analytical methods to obtain an absolute force estimate. It should be the aim of all sensor elements to obtain arterial information, so placing them closer can be considered for future design iterations.

Furthermore, in addition to sensor locations, the number of pressure elements should be further explored. Knowledge about a minimum number of sensor elements needed for accurate measurements is beneficial as it would minimize the sensor's complexity, power consumption, cost, and footprint. Additionally, collecting sensor characterization data that are more relevant to the actual application should be considered for future studies. For example, when worn, the sensor will be compressed constantly over a long period of time, which may lead to viscoelastic effects such as stress relaxation and creep of the elastomer, potentially affecting the validity of force estimates.

V. Conclusion

A sensor solution combining PPG and force detecting capabilities have been developed. The solution is constructed by off-the-shelf components that have a small footprint, low cost, and low power consumption. The force sensing is made possible by embedding piezoresistive barometric pressure transducers in soft silicone, protecting the sensor elements while communicating pressure changes, ensuring both high sensitivity and accuracy, in addition to a durable design. Preliminary tests show that the combined technology is able to monitor physiological information precisely, and it is hypothesized that the combined PPG/tonometry solution can be used to ameliorate one or more of the limitations associated with physiological measurements from the separate technologies. Therefore, the combined sensor prototype has the potential of giving current pulse waveform based blood pressure estimation models valuable additional inputs, ultimately increasing estimation accuracies and measurement confidence levels.

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