**Stretchable PPG sensor with light polarization for physical activity–permissible monitoring**

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Skin-attachable sensors, which represent the ultimate form of wearable electronic devices that ensure conformal contact with skin, suffer from motion artifact limitations owing to relative changes in position between the sensor and skin during physical activities. In this study, a polarization-selective structure of a skin-conformable photoplethysmographic (PPG) sensor was developed to decrease the amount of scattered light from the epidermis, which is the main cause of motion artifacts. The motion artifacts were suppressed more than 10-fold in comparison with those of rigid sensors. The developed sensor—with two orthogonal polarizers—facilitated successful PPG signal monitoring during wrist angle movements corresponding to high levels of physical activity, enabling continuous monitoring of daily activities, even while exercising for personal health care.

**INTRODUCTION**

The most popular form of wearable electronic devices is a wristband equipped with sensors for tracking the user’s health vitality. Generally, it consists of a photoplethysmographic (PPG) sensor, which senses the changes in blood volume associated with cardiac cycles. The PPG sensor itself—comprising light-emitting diodes (LEDs) to illuminate the skin and photodetectors to measure the amount of reflected or transmitted light—can acquire a wide range of information, including heart rate (HR), peripheral oxygen saturation, and respiration rate (1–3). However, a critical drawback of PPG sensors is the noise arising from physical motion artifacts. There are two primary reasons for such motion artifacts. The first is the changing contact between the sensor and skin, which can be identified using other sensors such as accelerometers and galvanometers and eliminated through data processing (4–7). However, for most wearables, discontinuities or distortions in the HR signals may occur when eliminating the motion artifacts from signals (8–12) if the HR signals are not notably stronger than those of the motion artifacts. One of the most widely studied methods for limiting changes in skin contact is the adoption of flexible and stretchable sensors in the development of skin-attachable sensors (13–19). These solutions render the sensor less sensitive to motion artifacts, as compared to rigid PPG devices. The second reason for motion artifacts is the temporal change in light scattering from the skin tissues owing to the changes in the relative position between the sensor and the skin, leading to inconsistent data. The distorted PPG signals caused by strong motion artifacts in existing wearables limit their use as medical and health care devices (20). Obtaining high-quality PPG signals during physical activities enables the diagnoses of various cardiovascular-related diseases (such as atherosclerosis) and heart diseases (such as arrhythmia and angina) (21) and also allows for the design of safer exercise rehabilitation programs. Real-time monitoring could aid in detecting the risk of heart disease and hypertension in advance and responding immediately.

Here, we demonstrate a skin-attachable PPG sensor with an orthogonal polarizer-analyzer (OPA) pair to reduce motion artifacts by suppressing the amount of light scattered from the skin epidermis before reaching the photodetector (Fig. 1A). The light emitted from the light source passes through the substrate, enters the skin beneath it, and is then scattered and reflected, and the photodetector subsequently receives this scattered/reflected light. In this study, the sensor was attached to the inner wrist, where skin movement is relatively large (fig. S1). The wrist was then flexed or extended periodically to varying degrees, thus changing the wrist angle, and the motion was evaluated quantitatively. Such wrist movement can cause structural changes such as the loss of contact between the sensor and skin, changes in the sensor geometry, and changes in skin tissue. Most of the light reflected from the skin is scattered at small depths from the skin surface, such as from filamentous proteins, collagens, or intracellular structures (Fig. 1B) (22, 23). This light reflected from skin tissues—excluding that reflected from blood vessels, which provides a constant signal during resting periods—causes a time-dependent variation in the signal with wrist movement and cannot be distinguished from the HR signals originating from blood vessels (Fig. 1C). When linearly polarized light is incident on the body, it is dispersed by scattering media such as human skin (24, 25). The polarization of the light reflected from small depths of the skin surface is biased with the initial direction. By contrast, the light reflected from blood vessels, located at a relatively greater depth, has sufficiently dispersed polarization (Fig. 1D). Consequently, the reflected light with polarization orthogonal to that of the incident light can be selected to suppress the light scattered by the skin epidermis. In this manner, the contribution of motion artifacts to the PPG signal can be reduced by over 90%, as compared to that of unpolarized light, for wrist angles ranging from $-10^\circ$ (flexion) to $+10^\circ$ (extension) (Fig. 1E).

**RESULTS AND DISCUSSION**

The intensities of the light reflected from skin were calculated using optical simulations based on the Monte Carlo ray-tracing technique.
The polarization dependency for suppressing reflected light is weakly dependent on the wavelength of the light source and the skin photoresponses. When linearly polarized light from an OLED with a polarizer is incident on the skin, the second orthogonally oriented polarizer (analyzer) cuts off the light scattered from small-depth tissues, which cause motion artifacts, while allowing the light from the deep blood vessels to pass through. Consequently, the light reflected from skin within a depth of 1.2 mm accounted for approximately 90% of the entire signal, corresponding to a direct current component of the PPG signal in the absence of geometrical changes. A fraction of the signal from the blood vessel—which could be a pulsatile alternating current signal corresponding to blood volume changes—accounted for less than 10% of the entire signal (26, 27). When linearly polarized light is incident on human skin, the light scattering caused by human tissues results in changes in the polarization state of the light. After a sufficient number of scattering events, the light eventually reaches an unpolarized (isotropic) state. When reflected from a depth of 1.0 mm or more in the human epidermis, light with a wavelength of 530 nm is completely unpolarized (Fig. 2A). Consequently, the light reflected from skin includes polarized signals from a depth of 1.0 mm, along with unpolarized signals from greater depths. Thus, by selecting the orthogonally polarized light, the amount of light scattered from a depth of 1.0 mm in the light reflected from skin can be selectively decreased. The polarization dependency for suppressing reflected light is weakly dependent on the wavelength of the light source and the skin photo-type (Figs. S2 and S3).

A polarizer-analyzer pair—in which the second polarizer is termed as an analyzer—is a general tool used to investigate polarization-sensitive light-matter interactions (28–30). A polarizer-analyzer pair can be used to select light with orthogonal polarization from a light source such as an organic LED (OLED) that emits unpolarized light. When light is transmitted by the polarizer and the adjacent analyzer located on the same plane with specular reflection, as shown in fig. S4, the light intensity, \( I \), is always zero if \( \alpha = 0^\circ \) or \( 90^\circ \) (see Materials and Methods). When considering light absorption and scattering at the skin, instead of specular reflection, the angles \( \alpha \) and \( \beta \) are dispersed by multiple scattering to ensure that the OPA does not completely block the light. In Fig. 2A, when light with orthogonal polarization relative to the initial direction is selected, the intensity of the light reflected from skin decreases by a maximum of 40%. Figure 2B shows the spatial distribution of light reflected from depths of 0.2 to 0.3 mm. The intensity of the reflected light without an analyzer, \( S_0 \), decreases exponentially with an increase in the distance from the light source along all radial directions. However, the light intensity with an analyzer, \( S_1 \), features anisotropic distribution, which represents a decrease in intensity along the \( x \) and \( y \) axes, corresponding to \( \alpha = 0^\circ \) or \( 90^\circ \) and presented in Fig. 2C. The intensity distribution of the reflected light from blood vessels is broad and isotropic in all radial directions and remains constant, regardless of whether an analyzer is present, because of the sufficient multiple scattering (Fig. 2, D and E).

In this study, a stretchable patch-type PPG sensor with an OPA was developed to achieve a mechanically stable lightweight system. An OLED was used as the light source and an organic photodiode (OPD) was used as the PPG signal detector (Figs. S5 and S6). The PPG sensor—consisting of two OLEDs and one OPD (Fig. 3A)—was rendered stretchable by using (i) an elastomer substrate, (ii) a stretchable cracked Au electrode, and (iii) an island-shaped high-modulus layer for stress relief near the device (31). The OPA was then embedded in the elastomer substrate. The distance between the OLED and OPD of the PPG device was carefully adjusted through optical simulations to minimize the motion artifact of the PPG signal and to maximize the heartbeat signal, thus obtaining high-quality PPG signals (Fig. S7). The patch-type PPG sensor had a signal-to-noise ratio (SNR) of more than 20 dB, comparable to that of a Si-based PPG sensor, though the signal intensity was reduced owing to the OPA, resulting in an SNR of 13 to 14 dB (Figs. S8 and S9). The detailed fabrication process of the stretchable sensor is provided in Materials and Methods.

The basic device characteristics of both the OLED and the OPD were investigated (Fig. S10). The OPD device exhibited high specific...
detectivity because of the moderate spectral responsivity, as well as a sufficiently low dark current density ($6.8 \times 10^{-10}$ A/cm$^2$) at a reverse bias of 1 V, a value comparable to those of commercial silicon-based photodetectors (table S2). Moreover, the OPD device was mechanically robust and could be operated stably under 30% strain (Fig. 3B). The OLED was also mechanically robust under 30% strain and exhibited a constant luminance of approximately 11,340 cd/m$^2$ with a current density of 17.74 mA/cm$^2$ at 6.0 V (Fig. 3C). A flexible printed circuit board was used as an interconnector for data acquisition from the PPG device attached to the inner wrist (Fig. 3A). To quantify wrist motion, the spatial profile of the skin surface around the radial artery was recorded based on changes in the wrist angle using a motion capture system (Fig. 3D and fig. S11). The range of wrist motion during the measurements varied from $-40^\circ$ (flexion) to $+60^\circ$ (extension) at 10° intervals, and the results obtained with different subjects were averaged. As shown in Fig. 3D, a tattoo patch consisting of black dots was attached to the inner wrist, and the movement of each dot was then observed using two cameras to quantify the movement of the skin surface using three-dimensional digital image correlation (DIC) analyses. In the range of the studied wrist angles, the maximum strain in the region near the radial artery was approximately 20%, which was within the mechanically robust range of the sensor (Fig. 3E).
The PPG signal was measured using three types of sensors, namely, a commercially available rigid sensor, stretchable patch-type sensor, and stretchable patch-type OPA sensor (fig. S12), located on the wrist at rest and during movement. Measurements were commenced at rest, and wrist movement was started after a few tens of seconds. The wrist was bent and stretched periodically to vary the wrist angle over a range of −20° to +20°. Periodic motion was realized at a frequency of ~0.5 Hz; this could be distinguished from the typical frequency of the HR, which ranges from 1 to 2 Hz. Both the rigid sensor and the patch-type sensor were affected by motion artifacts, whereas the patch-type OPA sensor clearly provided a consistent PPG signal (Fig. 4A). Through Fourier transform of the time-dependent signals obtained at rest and during wrist movements, the peak intensity at a frequency of 0.5 Hz—corresponding to the motion artifacts—was compared with that at a heartbeat frequency of 1.1 Hz (Fig. 4B). During the resting period, only the 1.1-Hz signal from the heartbeat and its harmonics signals were observed, whereas the signals due to the motion artifacts at 0.5 Hz appeared during wrist movement. The rigid and patch-type sensors exhibited stronger motion artifact signals than the heartbeat signal. The signals due to the motion artifacts for the rigid sensor and patch sensor were 3.3 and 1.8 times stronger than that of the HR, respectively. Evidently, the patch-type sensor was less vulnerable to motion artifacts than the rigid sensor. In the case of the patch-type OPA sensor, the peak intensity at 0.5 Hz was half the peak intensity at 1.1 Hz. The OPA sensor was able to suppress motion artifacts by 6.8 and 3.8 times, respectively, as compared to those of the rigid and patch-type sensors.

Figure 4C shows the variations in the ratio between the peak intensities at 0.5 and 1.1 Hz with a change in the wrist angle over the ranges of −1° to +1°, −5° to +5°, −10° to +10°, −20° to +20°, −30° to +30°, and −40° to +40°, respectively. It is clear that the patch-type sensor without the OPA produced a noisy signal due to motion artifacts, although it could maintain contact with the skin. The introduction of the OPA into the patch-type sensor resulted in a substantial decrease in the signal due to motion artifacts, with the noise level due to the motion artifacts (S_{0.5Hz}) remaining lower than the heartbeat signal (S_{1.1Hz}) up to a wrist angle of ±30°. The motion artifact suppression by the OPA was effective in the case of the sensor with red OLEDs (Fig. S13). The proposed patch-type OPA sensor exhibited an outstanding performance in comparison to that of the rigid sensor, which exhibited a higher noise level (owing to motion artifacts) than the heartbeat signal, even at ±1° (fig. S14). Moreover, the two patch-type sensors exhibited consistent levels of S_{0.5Hz} and S_{1.1Hz}, whereas the results obtained with the rigid sensor were strongly dependent on the adhesion strength of the sensor to the skin, imparted by the cover tape. The motion artifacts and the intensity of the heartbeat signal tended to change based on the range of the wrist angle (Fig. 4D). In the case of a discontinuous movement, the intensity changes in the heartbeat signal due to wrist movement may be treated as a motion artifact. Thus, the results suggest that the OPA could suppress both the motion artifacts caused by changes in the heartbeat signal and the motion artifacts directly generated by wrist movement. Furthermore, it was confirmed that the HR accuracy, as compared with that for an ECG chest strap, can be increased from 67.4 to 83.7% in daily activities such as walking and running by applying the OPA to commercialized smartwatches (fig. S18).

The degree of motion artifact suppression using the OPA was calculated by considering various geometric changes due to wrist movement. First, the surface angle difference between the OLED and OPD based on the possible positions where the sensor could be attached to the body—defined as the in-plane angle, θ, and out-of-plane angle, φ, (Fig. 5A)—was extracted through DIC measurements (fig. S15). For a wrist angle ranging from −40° to +40°, the in-plane angle was within 1.6°, whereas the out-of-plane angle tended to vary over a wide range (−3° to +12°). Figure 5B shows the predicted motion artifact corresponding to the ratio between the AC noise and the signal from the blood vessel with respect to the out-of-plane angle. The AC noise, N_{AC}, represents the difference in the intensities of light reflected from the epidermis at an out-of-plane angle...
between $0^\circ$ and $\phi$. When the out-of-plane angle was within $15^\circ$, the suppression of the $N_{\text{AC}}$ facilitated by the OPA was 2 to 11 times, whereas it was 1.3 to 1.5 times for the in-plane angle (fig. S16), compared with the $N_{\text{AC}}$ of the sensor without the OPA.

The effect of the distance between the OLED and OPD was also analyzed. The shorter the distance between the OLED and OPD, the greater is the intensity of the reflected light. Consequently, the light from the epidermis was more sensitive to the distance between the OLED and OPD than the light from the blood vessel (Fig. 2, B and D). For changes in the distance between the OLED and OPD from $-30\%$ to $+30\%$, corresponding to changes in the wrist angle from $-40^\circ$ to $+60^\circ$, the noise suppression using the OPA was found to be 1.3 to 1.6 times (Fig. 5C). Last, the effect of the variation in the adhesion strength between the skin and sensor was also evaluated, considering that the skin near the wrist exhibits the largest strain variation in the body. The conformal attachment of the skin patch to the rough and curvilinear surface of the skin is not optimal unless microstructures are used (32, 33). Although an adhesive elastomer was used as a substrate in the experiments, the gap between the skin and substrate could vary during wrist movement. The wider the gap, the stronger is the effect of the OPA. A noise suppression of 4.3 times was realized at a gap of 50 $\mu$m (Fig. 5D). It is worth noting that we experimentally confirmed the suppression of motion artifacts originating from the air gap using a conventional rigid OPA sensor (fig. S17).

To date, optical PPG sensors have played an important role in the development of health-monitoring devices and the growth in the market for wearable electronics. As the next generation of wearable devices, practical patch-type devices are expected to enable continuous health monitoring, even during intense physical activities. The approach of incorporating an OPA in a sensor offers new opportunities for effectively suppressing motion artifacts in the signals of optical sensors. In this study, the motion artifacts of a stretchable OPA sensor located on the wrist were successfully reduced more than 10-fold, as compared to those of rigid devices. The proposed method notably increases the permissible wrist angle range during wrist movement from less than 1° up to as much as 30°. We expect this OPA technology to improve the performance of future health-monitoring devices such as blood pressure sensors and glucose sensors, which rely on optical detection through the skin.

**MATERIALS AND METHODS**

**Transmittance when using two polarizers with specular reflection**

Assuming there is no light loss from the polarizer and analyzer, the intensity of the light transmitted by the analyzer is given by Malus’s law: $I = I_0 \cos^2 \theta$, where $I_0$ is the initial intensity of the light transmitted by the polarizer, and $\theta$ is the angle between the initial polarization...
direction and the polarization axis of the analyzer. When light is transmitted by the polarizer and the adjacent analyzer located on the same plane with specular reflection, as shown in fig. S4, the abovementioned equation can be modified as

\[ I = I_0 \cos 2\alpha \sin 2\beta \sin 4\beta / [(1 - \cos 2\alpha \sin 2\beta)(1 - \sin 2\alpha \sin 2\beta)] \],

where \( \alpha \) is the angle between the axis of the polarizer and the plane of light propagation, and \( \beta \) is the angle between the propagation direction of the incident light and the normal direction of the polarizer. Notably, the intensity of the light passing through the analyzer is strongly dependent on these angles, although the axes of the polarizer and analyzer are orthogonal with respect to each other. When \( \alpha = 0^\circ \) or \( 90^\circ \), the light intensity, \( I \), is always zero and does not depend on the value of \( \beta \). Consequently, the axes of the polarizers should be carefully adjusted on the basis of the position of the light source and detector in the adjacent polarizer-analyzer sensor.

**Stretchable substrate with stress-relief layers**

All the processing solvents, such as toluene, acetone, and 2-propanol (IPA), were purchased from commercial sources and used as received. The azide cross-linker bis(6-((4-azido-2,3,5,6-tetrafluorobenzoyl)oxy)hexyl)decadioate was synthesized from Medigen. The styrene-ethylene-butylene-styrene (SEBS) compounds H1043 and H1052, with poly(ethylene-co-butylene) (EB) volume fractions of 33 and 80%, respectively, were provided by Asahi Kasei. We used SEBS H1043 for the stress-relief layer (SRL) and SEBS H1052 for the stretchable substrate. Dextran, used as a water-soluble sacrificial layer, was purchased from Sigma-Aldrich and used as received.

A dextran solution [10 weight % (wt %) in deionized water] was spin-coated on an 8-inch glass wafer at 1500 rpm for 20 s. After baking on a hot plate at 180°C for 15 min, a 15-µm-thick SRL was fabricated on the dextran/glass wafer by spin coating a mixture solution of SEBS H1043 and azide cross-linker (100:5, w/w) in toluene (20 wt %) at 1000 rpm for 60 s. To remove the residual solvent, the substrate was baked at 120°C for 15 min. An SRL-patterning mask was then fabricated on the SEBS H1043 film, and selective areas were defined using a patterning mask exposed to ultraviolet (UV) light (wavelength: 254 nm) for 20 min to initiate the photocrosslinking reaction. Postbaking was conducted at 120°C for an additional 15 min on a hot plate in air to further increase the degree of cross-linking in the UV-exposed areas. The unexposed areas of the SEBS 1043 film were removed by washing with toluene two to three times, after which the SRL was baked at 120°C for 15 min. A 5-µm-thick stretchable substrate based on SEBS H1052 was then deposited on the SRL by spin coating a mixture solution of SEBS H1052 and azide cross-linker (100:3, w/w) in toluene (20 wt %) at 500 rpm for 60 s. After prebaking at 120°C for 15 min to remove the solvent, the SEBS H1052 film was exposed to UV light (wavelength: 254 nm) for 20 min to impart chemical resistance via the photocrosslinking of the SEBS film. Postbaking was conducted at 150°C for 15 min on a hot plate in air to further increase the degree of cross-linking.

**Microcrack Au interconnects**

GXR601 solution (as a positive photoresist) was photo-patterned onto the top of the stretchable substrate using a negative-tone photomask.
of Au interconnects. Au (100 nm thick) was thermally evaporated at a low deposition rate of 0.1 Å/s. The GXR601 layer was then repeatedly photo-patterned onto the Au-deposited layer using a positive-tone photomask of Au interconnects. Last, the Au layer on the photore sist was chemically etched using a gold etchant (Au10, UP chemical), and the residual photoresist on the Au layer was stripped using aceto nate and IPA to successfully obtain microcracked Au interconnects.

Stretchable PPG sensors

After the photo-patterning process of both the Indium Zinc Oxide (IZO)-based pixel and pixel-defined layer, the substrate was baked at 90°C for 20 hours to completely remove the trapped solvent. The substrates were loaded into a vacuum chamber for the thermal evaporation of the organic and metal layers under ca. 2 × 10⁻⁶ torr. Our green-OLED device has the following structure: SEBS/IZO (100 nm)/ HTL (100 nm)/host: green dopant (40 nm)/ETL (40 nm)/aluminum (160 nm). The green light-sensitive OPD device features a device structure of SEBS/IZO (100 nm)/HTL (100 nm)/boron subphthalocyanine chloride (SubPc) (22.5 nm)/ fullerene-C60 (C60) (45 nm)/bathocuprine (BCP) (10 nm)/Al (160 nm). For the thin-film encapsulation (TFE) process, 1.4-μm-thick amorphous perfluoro polymer solution (HyflonAD, Solvay) was coated on top of the PPG sensors. After baking at 70°C for 2 hours in a N2-filled glovebox, 10-mm-thick aluminum oxide (Al2O3) was deposited using the atomic layer deposition method. After the photo-patterning process on the TFE layer, the stretchable PPG sensors were peeled off the glass by immersing the entire device in deionized water to dissolve the sacrificial layer.

OPA film

To fabricate OPA film, three polarizer plates with dimensions of 1.1 mm by 1.1 mm were cut from three commercially available dichroic film polarizer for visible light, part number LPVISE2X2, Thorlabs Inc.). These were placed 1.9 mm apart on the SEBS film to correspond to the two OLEDs and an OPD and then pressed downward to ensure that they were embedded. The polarization direction of the central polarizer was disposed perpendicular to the polarization direction of the two polarizers on both sides. After the OPA film was peeled off from the glass substrate, it was laminated on the SEBS substrate of the patch-type sensor (Fig. 3A).

Characteristic measurements

Current-voltage (J-V) characteristics of the OPDs were measured using a Keithley K4200 semiconductor parameter analyzer. The external quantum efficiency spectra were obtained using a spectral photon-to-electron conversion efficiency measurement system, under monochromatic light generated by an ozone-free Xe lamp equipped with an optical filter and featuring a chopper frequency of 30 Hz. The intensity of the incident light was approximately 0.2 mW cm⁻² with an optical filter and featuring a chopper frequency of 30 Hz. Monochromatic light generated by an ozone-free Xe lamp equipped with a spectral filter and a chopper was used.

PPG measurements

Commercially available reflective PPG sensors (SFH7060, Osram Opto semiconductors Inc.) were used as the rigid-type devices. All the sensors (namely, the commercially available rigid sensor, stretchable patch-type sensor, and stretchable patch-type OPA sensor) were operated under the same measurement conditions with an evaluation module containing an integrated analog front-end 25 chip (AFE4490, Texas Instruments Inc.). The sampling rate was 500 Hz, and a notch filter with a frequency of 60 Hz was used to eliminate the resonance from the system as a noise source. No other algorithms were used for the devices.

PPG signal analysis

In the absence of motion artifacts, the SNR is calculated from the signal and noise in the frequency domain via Fourier transform of the time-voltage signal, as follows:

\[ \text{SNR} (dB) = 10 \log_{10} \left( \frac{\text{signal}}{\text{noise}} \right) = 10 \log_{10} \left( \frac{\int_{f_{s}/3}^{f_{s}} |I(\omega)|^2 \, d\omega}{\int_{f_{s}/3}^{f_{s}} |I(\omega)|^2 \, d\omega} \right) \]

Here, \( I(\omega) \) is the intensity of the power spectrum at frequency \( \omega \), the signal is the sum of the intensities in the range of 0.5 to 10 Hz, and the noise is the sum of the intensities in the range of 10 Hz to the maximum frequency. If motion artifacts exist, motion artifact suppression, \( R_m \), is calculated on the basis of the signal and the motion artifact in the frequency domain via Fourier transform of the time-voltage signal obtained from a photodiode, as follows:

\[ R_m = \frac{\max(\{I(\omega)\}, 0.3-0.7Hz)}{\max(\{I(\omega)\}, 0.9-1.3Hz)} \]

Optical skin modeling

Optical skin modeling was performed using the commercially available LightTools software based on the Monte Carlo ray-tracing technique. Multilayered human tissues were simplified to two or three layers comprising the epidermis/dermis, blood, and subcutaneous fat. The detailed scattering and absorption coefficients used for the simulations are presented in table S1. To simplify the light-tissue interactions, we assumed homogeneous scattering media and used the angular distribution of light scattering based on the Heneyy-Greenstein phase function. Over 10 million rays were traced for a detailed skin depth profile.

Simulation of motion artifacts

Following skin modeling with the light source and detector corresponding to the size and distance of the OLED and OPD in the sensor, the intensity of light reflected from the entire skin (including the dermis, vessels, and fat), \( I_{\text{total}} \), the intensity of light reflected from the dermis (including the dermis and vessels), \( I_{\text{vessel + dermis}} \), and the intensity of light reflected from the dermis, \( I_{\text{dermis}} \), were simulated. Light reflected solely from the blood vessels, corresponding to the heartbeat signal, is calculated as \( I_{\text{vessel}} = I_{\text{vessel + dermis}} - I_{\text{dermis}} \). Light reflected from the non-vessel skin is defined as \( I_{\text{non - vessel}} \). The motion artifact suppression achieved by the OPA was quantified as the ratio of \( N_{\text{AC, OPA}} \), calculated with two polarizers, and \( N_{\text{AC, unpol}} \), calculated without the two polarizers. All the volunteers confirmed their ethical approval by signing an informed consent before participation in this study. On the basis of the Enforcement Rule of Bioethics and Safety Act, from the Korean Ministry of Health and Welfare, the authors confirm that all the experiments performed on human subjects were not subject to local ethics committee approval because there was no invasive measurement or physical modification on humans.
REFERENCES AND NOTES

1. K. H. Shelley, Photoplethysmography: Beyond the calculation of arterial oxygen saturation and heart rate. Anesth. Analg. 105, 531–536 (2007).

2. J. Allen, Photoplethysmography and its application in clinical physiological measurement. Physiol. Meas. 28, R1–R39 (2007).

3. L.-G. Lindberg, H. Uggeln, P. Å. Öberg, Monitoring of respiratory and heart rates using a fibre-optic sensor. Med. Biol. Eng. Comput. 30, 533–537 (1992).

4. H. H. Asada, H.-H. Jiang, P. Gibbs, Active noise cancellation using MEMS accelerometers for motion-tolerant wearable bio-sensors, in Proceedings of the 26th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (San Francisco, CA, USA, 2004), pp. 2157–2160.

5. S. H. Kim, D. W. Ryoo, C. Bae, Adaptive noise cancellation using accelerometers for the PPG signal from forehead, in 2007 Proceedings of the 29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (2007), pp. 2564–2567.

6. R. W. C. R. Wijshoff, M. Mischi, J. Veen, A. M. van der Lee, R. M. Aarts, Reducing motion waves by spherical diffusers: Influence of the size parameter. Phys. Rev. E 49, 1767–1770 (1994).

7. J. Lee, M. Kim, H.-K. Park, I. Y. Kim, Motion artifact reduction in wearable photoplethysmography based on multi-channel sensors with multiple wavelengths. Sensors 20, 1493 (2020).

8. A. Scherbina, C. M. Mattsson, D. Waggott, H. Salisbury, J. W. Christie, T. Hastie, M. T. Wheeler, E. A. Ashley, Accuracy in wrist-worn, sensor-based measurements of heart rate and energy expenditure in a diverse cohort. J. Pers. Med. 7, 3 (2017).

9. B. Bent, B. A. Goldstein, W. A. Kibbe, J. P. Dunn, Investigating sources of inaccuracy in wearable optical heart rate sensors. npj Digit. Med. 3, 18 (2020).

10. M. Etiwy, Z. Akhrass, L. Gillinov, A. Alashi, R. Wang, G. Blackburn, S. Gillinov, D. Phelan, A. M. Gillinov, P. L. Houghtaling, H. Javadikasgari, M. Y. Desaid, Accuracy of wearable heart rate monitors in cardiac rehabilitation. Cardiovasc. Diagn. Ther. 9, 262–271 (2019).

11. M. P. Wallen, S. R. Gomersall, S. E. Keating, U. Wislaff, J. S. Coombes, Accuracy of heart rate watches: Implications for weight management. PloS ONE 11, e0154420 (2016).

12. S. Gillinov, M. Etiwy, R. Wang, G. Blackburn, D. Phelan, A. M. Gillinov, P. Houghtaling, H. Javadikasgari, M. Y. Desaid, Variable accuracy of wearable heart rate monitors during aerobic exercise. Med. Sci. Sprots Exerc. 49, 1697–1703 (2017).

13. J. Kim, G. A. Salvatore, H. Araki, A. M. Chiarelli, Z. Xie, A. Banks, X. Sheng, Y. Liu, J. W. Lee, K.-I. Jang, S. Y. Heo, K. Cho, H. Luo, B. Zimmerman, J. Kim, Y. Lan, X. Feng, S. Xu, M. Fabiani, G. Gratton, Y. Huang, U. Paik, J. A. Rogers, Battery-free, stretchable optoelectronic systems for wireless optical characterization of the skin. Sci. Adv. 2, e1600418 (2016).

14. T. Yokota, P. Zalar, M. Kaltenbrunner, H. Jinno, N. Matsuhisa, H. Kitanosako, Y. Tachibana, C. Newsome, R. Wilson, A. C. Arias, A flexible organic reflectance oximeter array. Adv. Healthc. Mater. 5, 80–87 (2016).

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