A motion artifact generation and assessment system for the rapid testing of surface biopotential electrodes

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Abstract
Dry electrodes can reduce cost while increasing the usability and comfort of wearable monitoring systems. They are, however, susceptible to motion artifacts. The present electrode testing methods lack reliability and do not separate the factors that affect the motion artifact. In this paper, we introduce a first generation motion artifact generation and assessment system that generates the speed, amplitude, and pattern-wise programmable movement of the electrode. The system simultaneously measures electrode–skin impedance, the motion artifact, and one channel of an electrocardiogram that contains the motion artifact and monitors the mounting force applied to the electrode. We demonstrate the system by comparing the applied movement and the measured signals for electrode movements up to 6 mm and movement frequencies from 0.4 Hz to 4 Hz. Results show that the impedance change and surface potential are visually clearly related to the applied motion, with average correlations of 0.89 and 0.64, respectively. The applied force, electrode location, and electrode structure all affect the motion artifact. The setup enables the motion of the electrode to be accurately controlled. The system can be used as a precursor to the testing of integrated systems because it enables thorough, repeatable, and robust motion artifact studies. The system allows a deeper insight into motion artifacts and the interplay of the various factors that affect them.

Keywords: dry electrodes, surface electrodes, ECG, EMG, motion artifact, electrode-skin impedance, testing, prototyping

(Some figures may appear in colour only in the online journal)
1. Introduction

In this work, we describe the design for a device that enables the measurement of the motion artifact of surface electrodes in response to controlled and standardized motion under a controlled environment. The motion artifact is an important challenge that affects the use of dry electrodes in applications that involve the measurement of surface biopotentials such as electrocardiography (ECG), electroencephalography, electromyography (EMG), and the electro-oculogram (EOG).

ECG is the main tool used in the diagnosis of cardiovascular disease, and EMG is widely used in physical rehabilitation and athletics (Clarys 2000, Lamontagne 2001). With the decrease in cost and size of electronics, previously laboratory-bound equipment is making its way onto the training field and into the daily life of the population in the form of wearable monitoring systems. In the long-term, the cost-effective and user-friendly use of these methods in application-specific conditions, such as everyday activities or athletic training, ensure that reusable electrodes are a practical solution.

The electrodes that are used in clinical, wearable monitoring systems need to work reliably over long monitoring periods. Furthermore, the electrodes must be user-friendly, cause no discomfort to the wearer, and ideally be user-applicable. Electrodes that are made from conductive textiles, polymers, rubber, silicon, or other composite materials that have suitable electrical properties are washable, comfortable, and reusable, and thus fulfill the implementation criteria. The connection of these electrodes to the electronics using textile or polymer cables integrated into the clothing will enable the wearable systems to make leeway into multiple usage areas, and possibly create new applications.

These types of electrodes are attached onto the skin without using conductive paste or glue, and are most commonly attached with tight elastic clothing, bands, or ribbons. These electrodes are called dry electrodes because of the lack of conducting gel or paste between the skin and the electrode. During use, however, the skin sweats under the electrode and forms a moist layer that increases the conductivity of the electrode–skin interface (Gruetzmann et al 2007). In this paper, therefore, we use the terms ‘textile electrode’ and ‘dry electrode’ interchangeably, even if the electrodes are moist or wet. Because of the lack of a stabilizing gel layer at the electrode–skin interface, the lack of glue used to keep the electrode firmly in place, and the high electrode–skin impedance in the case of dry skin, dry electrodes tend to be especially prone to the effects of motion (Meziane et al 2013). As a result, such electrodes are not used in medical mobile systems.

Dry electrodes have, however, been successfully implemented in heart rate monitors made by companies such as Polar (Polar Electro, Kempele, Finland) and Suunto (Suunto, Vantaa, Finland) for sports applications, where the pulse is the main parameter of interest.

In the design of dry electrodes, there are many parameters that can be optimized. These parameters include the electrode material, the electrode surface texture that affects the electrode–skin interface structure, the design of the electrode component that is located between the electrode surface and the electrode mount system, and the force applied to the electrode by the mount system (Comert et al 2013).

Webster and Ödman state that the most prominent cause of motion artifacts is the change in the potential between the outer and inner layers of the skin (Tam and Webster 1977, Ödman and Öberg 1982). The outer layer of the skin is more negative than the inner layers and this potential difference across the epidermis is attributed to so-called injury currents (de Talhouet and Webster 1996). The skin deformation that causes this potential change can arise from the lateral stretching of the skin, rotational stretching of the skin, or vertically applied pressure. This deformation also causes a change in skin impedance, but the potential changes do not
depend on this impedance change (Ödman and Öberg 1982). The input impedance of the bio-potential amplifier and the common mode rejection ratio determines the effect of these factors on the measured potentials. The explanation given by Webster for gelled electrodes might be inadequate for dry electrodes that rely on a thin layer of skin moisture. This moisture layer does not provide a continuous contact area. Variations in this layer, such as non-contact areas, might occur as a result of motion, posture, or less than optimal electrode design (Medrano et al. 2007). Dry electrodes are also prone to changes in the ionic concentration near the electrode surface that create a potential difference between the electrode and the electrolyte (Neuman 1997). Due to these factors, the notion that electrode–skin impedance is not so important in the generation of the motion artifact might be inadequate in the absence of conductive gel. Because the motion artifact of dry electrodes is not only caused by electric changes in the skin but also by the contact area and impedance changes in the conductive layer between the electrode and the skin, which is especially affected by sweat, these types of electrodes are more prone to motion artifact compared to standard medical electrodes (Comert et al. 2013).

Motion artifact studies that attempt to understand and eliminate the problem use various methods to create motion artifacts. Some methods can include having the subject walk or run on a treadmill (Ottenbacher et al. 2008b), carry out predetermined multiple joint limb movements (Mühlsteff et al. 2004, Kearney et al. 2007, Cho et al. 2009), predetermined single joint movements (Comert et al. 2013), or hardware-guided movements (Roy et al. 2007). Other methods include the pressing and moving of the electrode by an operator (Tong et al. 2002, Buxi et al. 2012), increasing the force over an electrode and monitoring the pulse response (Beckmann et al. 2010), and the measurement of motion artifacts caused by vibration (Searle and Kirkup 2000, Fratini et al. 2009). These methods, with the exception of the last two, are constrained by the human factor. While the movements will be fairly similar on the whole and will likely converge to an average form, differences exist between subjects, and even for the same subject between experiments. Furthermore, the methods can take time to set up, the repeated motion can be tiring for the subject, and the motion can be altered.

In a previous experiment, we proposed to synchronize the movement to a metronome with a clear definition of the range (Comert et al. 2013). We noticed, however, that there is room for improvement such as in controlling the motion, isolating the various factors that affect the motion artifact, and reducing the experiment set-up times.

The previous studies using dry electrodes were carried out with the electrode integrated into a garment (Cho et al. 2009, Buxi et al. 2012, Liu et al. 2013). As a result, the motion artifact was influenced by the design of the electrode mounting system, the interaction of the electrode with the mounting system, and the interaction of the skin with the electrode. The controlled generation of the motion artifact directly on the electrode enables the factors related to the motion artifact to be studied more independently, before the electrode is integrated into a wearable system.

Moreover, a method is needed that can provide controllable movements with various frequencies and magnitudes for a thorough investigation of the electrode behaviour over a variety of motion parameters, under controlled conditions, that can be repeated on a number of test subjects and electrode designs.

Such methods have been previously proposed by Ödman et al. (Ödman and Öberg 1982) and latterly by Liu et al. (2013). Both groups used motorized systems that created motion at the electrode and measured the impedance between the electrode and the medium and the potential observed at the electrode. Ödman et al. studied the back and forth motion of a gel-based electrode on the skin using a stepper motor and, monitored the electrode–skin impedance and the surface potential, as well as visually tracking the motion with a stroboscope and camera. Liu et al. studied the electrode movement on a skin-simulating membrane by measuring
the electrode–membrane impedance and the resulting potentials, and by monitoring the force applied on the electrode.

Studies looking at the relationship between electrode motion and changes in the electrode–skin interface impedance have found that the impedance changes with applied motion. These studies have been done at various impedance current frequencies: 7 frequencies from 120 Hz to 1.8 kHz (Hamilton et al 2000), 200 Hz and 2 kHz (Ödman and Öberg 1982), 400 Hz (Ottenbacher et al 2008a), 2.2 kHz (Buxi et al 2012), 100 kHz (Comert et al 2013) and a square wave at 2 kHz (Romero et al 2011).

In this paper, we introduce a versatile and affordable, partially 3D-printed, controllable electrode motion generation and motion artifact assessment system that can be used in various locations on the body. The system comprises a programmable horizontal motion generation system combined with a force sensor for monitoring vertical forces and a previously developed bioelectric measurement setup (Comert et al 2013) that has been integrated into it. The assessment system allows for the simultaneous measurement of artifact free ECG, ECG free motion artifact and ECG affected by motion artifact together with real time electrode–skin impedance. We describe the entire system and demonstrate its functionality by presenting sample data and investigating the relationship of the ECG, the motion artifact and the electrode–skin impedance to the known motion; while also monitoring the force applied on to the electrode. We test two textile electrode designs to investigate the system’s ability to show differences in electrodes. Two locations are tested to further understand the capabilities of the system.

2. Methods

2.1. System overview

The developed motion artifact generation and assessment system is comprised of two parts: the motion generation module and the biosignal measurement module. By moving the electrode back and forth in respect to the skin using the motion of the generation module, the motion artifact is created. The resulting signals are then measured by the biosignal measurement module and stored and displayed on a PC connected to the modules via USB. The system overview, which is explained below, is presented in figure 1(a).

2.2. Motion generation module

The motion of the electrode is created by a servo that is attached to a platform that can be moved vertically to the skin. This platform is lowered to mount the electrode onto the skin, and a force sensor, located in a housing between the servo and the electrode, measures the mounting force. A microcontroller board controls the servo and also processes the information about the mounting force obtained from the force sensor. This motion generation system, which is described in detail below, is presented in figures 1(b) and (c).

An Arduino UNO microcontroller board (SmartProjects, Turin, Italy) is used to control the servo and monitor the mounting force. A PC was used to program the microcontroller via USB. Once programed, it is a standalone system. Upon receiving the appropriate signals from the BioPac system, which is introduced in the data measurement module section below, the microcontroller starts and stops the motion.

A Dremel 220 workstation (Robert Bosch Tool Corp., Illinois, USA) forms the vertically moving platform for the servo-force sensor-electrode combination. The platform has been
modified into a manually operated mounting system so that the lever handle of the workstation can be used to move the platform to press the electrode onto the skin. The more the lever handle is pulled down and the platform lowered, the more force is exerted onto the electrode, and thus the electrode applies more pressure to the skin. The ratio of lever handle movement to platform movement is 176° of lever rotation for 5 cm of platform movement. This also can be written as 43 cm of travelled distance of the tip of the lever handle for 5 cm travel of the platform. This high ratio of lever arm distance to platform movement distance and the mechanical friction and displacement tolerance ensure that even if the lever has to be held in the same location manually, the operator can keep the platform steady during the experiment, and ensures steadily applied mounting force on the electrode. The Dremel has a metallic base, and the Dremel was located on a wooden desk, with no part of the workstation connected to the mains, other device, or other person.

The servo that generates the motion is a hexTronik HX12K Hi-speed servo (Hextronik Limited, ChenDu, China). The high torque of the servo ensures smooth operation under load. The electrode is connected to the servo arm via force sensor housing. The rotation of the servo axis causes the rotation of the servo arm, and this causes the movement of the electrode. The servo is controlled by a pulse width modulated signal, which is provided by the microcontroller board that specifies the target angle of the servo axis. The movement of the electrode is achieved by increasing or decreasing the target-angle with every consecutive command.
resulting in a semi-continuous rotation of the servo arm. By changing the difference between the target angles of the consecutive commands, the speed of the servo is changed.

The back and forth movement is circular in nature, but at the angles used the angular effect is small and can be neglected so as to assume movement is linear. Equation (1) gives the relationship between the rotation of the servo and the displacement of the electrode for the $x$ axis and equation (2) for the $y$ axis. The servo arm is 25 mm in length and $\Delta \text{deg}$ is the angle of servo rotation from the starting point to the end of the rotation range. Visually, this relationship is shown in figure 2(a).

$$\Delta x = 25 \times \sin \Delta \text{deg}$$  \hspace{1cm} (1)  

$$\Delta y = 25 \times (1 - \cos \Delta \text{deg})$$  \hspace{1cm} (2)

As can be seen, the speed of the rotation and the extent of the rotation can be modified. Changing these parameters also controls the frequency of the motion. Keeping the angular difference between consecutive commands constant, the angle versus time relationship of one back and forth motion we applied to the electrode has a triangular pattern. As per equations (1) and (2), this results in a slight concave slope for the $x$-axis movement and a slight convex slope for the $y$-axis movement of the electrode. By changing the difference between the target angles of consecutive commands, different patterns such as a sinusoidal-like pattern of the angle versus time relationship can be achieved.

The execution of each command cycle, which includes the time the servo requires to reach the target angle and the time needed by the Arduino UNO microcontroller to execute the code, takes approximately 62.5 milliseconds (ms). This execution time includes some error checking and the display of information, and in an automated system can be reduced considerably. The servo refresh time between two consecutive commands is a minimum of 50 ms, limiting the servo back and forth rotation frequency to 10 Hz. The rotation speed of the used servo under no load is 300 degrees/s. As an example, for this specific servo, this limits the rotation magnitude of a 10 Hz back and forth movement to 15°. It needs to be noted that the electrode displacement not only depends on the rotation angle, but also the length of the servo arm transforming the rotational movement into the lateral motion of the electrode.
A FlexiForce A201 force sensor (Tekscan Inc., South Boston, USA) was used to measure the pressure/force applied to the electrode to mount it on the skin. The FlexiForce A201 is a thin film force sensor with a circular sensing area with a diameter of 0.375 in. that is essentially a non-linear force sensitive potentiometer. The resistive output of the force sensor was linearized by the use of an inverting Op-Amp circuit as recommended by the user manual of the sensor, whose voltage output is fed to the microcontroller board. After calibration, the sensor is linear to ±3%.

The force sensor is located in a three-piece enclosure that is located between the servo arm and the electrode. The top piece, which is connected to the servo arm, comprises a 3D printed plastic part that serves as a housing for the force sensor and an inserted Kevlar tube to which the bottom piece is inserted. The bottom piece comprises a 3D printed design that acts as a base for the electrode to attach to, and a cylinder that is inserted into the top part and is located exactly on the force sensor acting as a puck. Using Kevlar for the intermediate tube ensures low vertical friction, and the dimensions of the tube and the inserted cylinder ensure the secure placement of the electrode platform into the force sensor housing, as depicted in figure 2(b). The restriction of the lateral movement of the cylinder puck in the Kevlar tube ensures the minimization of shear forces on the sensor, forces which could cause errors in the measurements and reduce the life of the sensor.

At the selected location, the force sensor monitors the vertical force exerted on the electrode by the mounting system. The pressure between the electrode and skin is then the force divided by the electrode area. The monitored force translates to the force exerted on the electrode by the wearable garment. The location for the force sensor is chosen because of the difficulty of reliably measuring the pressure between the electrode and skin, due to the uneven shape of skin and underlying tissue and the changes in these layers caused by movement.

Three LEDs located on the moving platform are controlled by the Arduino microcontroller to inform the operator of the applied force. If the modified output of the FlexiForce A201 sensor, fed to the Arduino, is inside a target range the appropriate LEDs are lit up. For any target force level, the range is defined as the target force plus/minus 100 g. Having the LEDs lit up and located in an easily seen location provides easily accessible feedback and allows the force to be kept in similar ranges in between experiments and with different subjects, further aiding the standardisation and repeatability.

Before the experiments, the FlexiForce A201 sensor needed to be primed for the experiment as stated in the user guide. A soft padding was pressed onto a Soehnle Siena (Leifheit AG, Nassau, Germany) kitchen scale with an accuracy of 1 g for a few minutes at 110% of the highest target force level to be used in the experiment. The output given by the Arduino after this time was used to program the calibrations for the LED control parameters back into the Arduino.

In the present design, motion is created by laterally moving the electrode on the skin, creating deformation of the skin directly under the electrode. Skin stretch can be caused by parallel movement in the form of linear displacement, or parallel movement in the form of rotational displacement, or due to the perpendicular displacement of the skin, and their effects on the potential generated are assumed to be similar (Ödman and Öberg 1982). For the purposes of this study, we focused on the motion artifact created by lateral movement. We monitored and controlled the vertical pressing force on the electrode to prevent any motion artifact caused by vertical skin stretch.

2.3. Biosignal measurement module

The BIOPAC data acquisition system (BIOPAC Systems, Inc., California, USA) was used for measuring the biosignals. The MP35 data acquisition unit was used to measure one ECG
channel approximating the standard ECG channel aVR and one biopotential channel of almost purely motion artifact. The EBI100c bioimpedance module was used for an electrode–skin impedance measurement at 100 kHz input current frequency, using a three electrode setup, with the sensitivity field maximized under the electrode which is subject to motion (Comert et al 2013). Even with a higher frequency than most biopotential measurements for which motion artifact is of importance, impedance measurements done at this frequency accurately show the effect of electrode motion on the skin–electrode interface and on the skin layers. The setup of these channels is shown in figure 3(a), and explained below. The analog output of the BIOPAC was used to start and stop the motion in a programmed time window. The BIOPAC MP35 data acquisition unit has an input impedance of 2 MΩ. At the used settings, the MP35 unit has a resolution of 3 µV, and the EBI100c bioimpedance module has a resolution of 0.0015 Ω and measures the impedance using a 100 µA current.

Ch2 measured ECG affected by motion, between electrodes ch2+ and ch2−. Ch1 measured the motion artifact with negligible ECG interference between ch1+ and ch1−. As shown in figure 3(a), the negative potential measurement point for these two channels, namely ch1− and ch2−, were set to be the electrode subjected to motion. When the electrode is moved the potential at the electrode changes and since the other electrodes are steady this is picked up by the two channels as motion artifact. Electrode–skin impedance was measured simultaneously with the biopotential measurements from a three electrode impedance measurement setup. The current source and positive voltage nodes were set as the electrode subjected to motion, picking up any impedance change occurring at the electrode location, so that all three measurements, ECG, motion artifact, and impedance were affected by the same electrode movement. The current sink and negative voltage electrodes of the impedance measurement (side of the upper arm and inner palm, respectively) were separate from the positive potential electrodes for the surface biopotential measurements (on the ribcage under the nipple for ch2 and back of the hand for ch1).

The electrodes affected by the motion were designed as dry electrodes made of textile. The electrode material was MedTex P180 (Statex Productions & Vertriebs GmbH, Bremen, Germany) warp knitted textile. The textile is made of highly conductive silver coated yarn, and is suitable for surface biopotential measurements. During separate experiments, the textile electrodes were applied to two different locations on the inner forearm to investigate

Figure 3. (a) The electrode locations: location 1 and 2 are the locations of the electrode that are subjected to motion in separate experiment sessions. Location 1 is on the inner forearm, approximately 7 cm proximal from the wrist, while location 2 is on the inner forearm, approximately 10 cm distal from the crease of the elbow. (b) The electrodes used in the studies.
the effects of the underlying tissues on the functioning of the system. The first location was approximately 7 cm proximal from the wrist on the inner forearm, while the second location was also on the inner forearm, approximately 10 cm distal from the crease of the elbow. An approximate depiction can be seen in figure 3(a). Both locations were selected so that they were on the same side of the inner arm as the palm. The four electrode leads for this electrode were connected to snap connectors which were soldered to one female snap connector. This female snap connector was then attached to the male snap connector that was part of the electrode structure and connected to the conducting textile of the electrode. Four drops of tap water were applied to the electrode to simulate the presence of sweat. This layer acts as an ionic conductor, lowering the skin–electrode interface impedance and improving the electrical contact. The use of this conductive layer is based on findings that for experiments of longer duration, perspiration of the skin under the electrode, which starts a few minutes after the mounting of the electrode, acts like a conductive electrolyte layer (Meziane et al 2013).

As the moisture at the electrode and the impedance of the skin–electrode interface remained fairly stable during the experiments, an intermediate application of water was not found to be necessary. Ambu Blue Sensor P electrodes (Ambu A/S, Ballerup, Denmark) were used at the locations not affected by motion and connected to the measurement system via standard electrode leads. These locations are also shown in figure 3(a).

Three different electrode designs were used as the motion affected electrode to investigate if the system could differentiate between the effects of different electrode designs. The first two were textile electrodes, with a 25 mm diameter, supported by a Poron XRD impact protection foam padding (Rogers Corporation, Rogers, USA). The first electrode had 50 mm diameter foam padding on top of the electrode and the second had foam padding the same size as the electrode. The thickness of the foam padding was 4 mm. The third electrode was an Ambu Blu electrode pressed on the skin via 25 mm diameter foam padding. These electrodes are shown in figure 3(b). The skin was not prepared other than the application of water.

2.4. Experimental procedures

The testing was carried out in two stages. The first stage was to investigate and gain an understanding of the general functioning of the system and the effect of the parameters related to motion. The second stage entailed a more detailed look at the effects of applied force on the electrode and to investigate the possible effects of electrode support structure design.

2.4.1. Stage 1. The aim of this stage was to test the capabilities of the motion generation system in creating the motion artifact and to study the effects of three different changes obtained by varying the basic motion parameter’s speed and rotation magnitude: increasing the rotation amplitude while keeping the speed constant, increasing the speed while keeping the rotation constant, and simultaneously increasing both the speed and the rotation amplitude while keeping the frequency of the movement constant. An approximation of these rotation patterns is depicted in the top row of figure 4. In this initial phase, the testing was carried out with one healthy male subject at two different electrode locations: the distal inner part of the forearm and the proximal inner part of the forearm, as shown in figure 3(a). One subject was assumed to be enough because even if there are skin tissue variations from person to person, and even between electrode locations for the same subject, the basic effect of motion creating the motion artifact is seen throughout the population. The two locations were, therefore, used to counter the disadvantage of having only one subject by creating the motion artifact in locations with different tissue parameters.
One experiment session comprised 3 blocks of 20 s of back and forth motion. Each block had a different change imposed on the parameters of the electrode motion generated by the rotational motion of the servomotor, which was seen as the angle of the servo arm versus time.

In the first type of motion pattern (pattern 1), we increased the range of motion throughout the motion period while keeping the rotation speed of the servo constant. The range of motion was increased from 2° (0.9 mm) to 12° (5.2 mm) while the speed was kept constant at 16 degrees/s (~6.9 mm s⁻¹). This meant that the frequency of the movement decreased from 4 Hz to 0.67 Hz.

In the second type of motion pattern (pattern 2), we increased the rotation speed while keeping the range of motion constant. The range of motion was set to 10° (4.3 mm) and the initial rotation speed was set to 8 degrees/s (~3.5 mm/s). During the experiment, the speed was gradually increased to 40 degrees/s (~17.5 mm/s) in the given rotation range and thereby increased the frequency of the movement from 0.4 Hz to 2 Hz for the same electrode displacement.

In the third type motion pattern (pattern 3), we increased the speed and the angle by the same factors. The starting range of motion was 2° (0.9 mm) and this was gradually increased to 14° (6 mm), while the rotation speed was increased from 8 degrees/s (~3.5 mm/s) to 56 degrees/s (~21 mm/s). This meant a movement where the frequency stayed the same at 2 Hz for a larger displacement of the electrode after each change.

How these rotational patterns translate into linear displacements in the xy plane in millimetres is depicted in figure 4.

For the textile electrode supported by the larger 50 mm diameter foam padding for both locations, three such sessions were carried out for each of the two different pressing forces of 750 g and 1250 g. Using Ambu Blu electrodes for both locations, one session was carried out for each of the applied force levels.

Figure 4. The three different movement patterns used in stage 1. Two peaks for each motion sequence are plotted. The upper row is the angular change, while the lower row shows the x dimension displacement in green and the y dimension in red. Pattern 1 implements and increases the angle. Pattern 2 implements and increases the speed. Pattern 3 implements and increases both angle and speed. The units and scale for the y-axes are the same from left to right, and for the x-axes they are the same from top to bottom.
This resulted in 12 sessions with textile electrodes and 4 sessions for commercial gelled electrodes, as shown in Table 1. Each session consisted of one set of data for each of the three movement patterns.

### 2.4.2. Stage 2

A second experiment was run to demonstrate and investigate in more detail the effect of applied force by applying five different forces on the electrode and to investigate the ability of the system to show different behaviours between different electrode designs. The selected target forces were 250 g, 500 g, 750 g, 1000 g, and 1250 g. The two electrodes tested were textile electrodes, one with a 25 mm diameter support padding and the other with a 50 mm diameter support padding, made of the same material.

The electrode set-up and main system behaviour and preparation were kept the same, but only the distal inner forearm location was used to prevent the factors related to differing tissue properties from affecting the experiment.

The motion protocol was changed so that only two different motion patterns were used. The two motion patterns were increasing the speed for the same range of motion (pattern 2), and increasing the speed coupled with increasing the range of motion (pattern 3). Each pattern lasted for 25 s and, together with a 1 s pause between them, formed one experiment session that lasted for 51 s. The first motion pattern was set to increase speed for a movement range of a steady 10° (4.3 mm). The starting speed was 8 degrees/s (~3.5 mm s⁻¹), and this was eventually increased to 48 degrees/s (~21 mm s⁻¹). Over a period of 25 s, the frequency increased from 0.4 Hz to 2.4 Hz. This was similar to motion pattern 2 of stage 1 (shown in figure 4), with one extra level of speed increase.

After the first motion, a consecutive 1 s pause was added, and then the second motion pattern was started. This was similar to pattern 3 of stage 1 (shown in figure 4) where the speed and the angle were increased simultaneously, thereby keeping the frequency of motion steady at 1.6 Hz. The initial range of motion and speed of 2.5° (1 mm) and 8 degrees/s (~3.5 mm s⁻¹) were increased to 12.5° (5.4 mm) and 48 degrees/s (~21 mm s⁻¹), respectively.

Six repeats of the five consecutive sessions for increased applied force were carried out for each of the textile electrodes. Table 1 depicts these procedures.

### 2.5. Data analysis

Since we were interested in motion artifacts with a minimum base frequency of 1 Hz, corresponding to a slow walk, the obtained data was band-pass filtered at 0.4–40 Hz. With a lower

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**Table 1. Summary of the experiments.**

| Electrode types (number of repeats per electrode per location per force) | Stage 1 | Stage 2 |
|---------------------------------------------------------------|--------|--------|
| 50 mm Padding (3 repeats)                                      | 50 mm Padding (6 repeats) |
| Ambu Blu (1 repeat)                                            | 25 mm Padding (6 repeats) |
| Pattern 1                                                     | Pattern 2 |
| Pattern 2                                                     | Pattern 3 |
| Pattern 3                                                     |        |

| Applied motion patterns | Pattern 1 | Pattern 2 |
|-------------------------|-----------|-----------|
| Pattern 2               | Pattern 3 |

| Mounting force levels   | 750 g     | 250 g     |
|-------------------------|-----------|-----------|
| 1250 g                  | 500 g     |
| 1000 g                  | 750 g     |
|                         | 1250 g    |

| Electrode locations     | Distal Forearm | Distal Forearm |
|-------------------------|----------------|----------------|
| Proximal Forearm        |                |                |
cut-off frequency of 0.1 Hz, there was still some low frequency drift left in the motion artifact. Because this signal was not caused by the motion of the electrode, generated by the servo, that was the scope of this study or by changes in the applied force, we used a 0.4 Hz lower cut-off high pass filter.

Power spectrum densities (PSD) of the ECG, motion artifact, and impedance signals were calculated, and normalized to be between 0 and 1. Because of this normalization, the presented PSDs have no units, and do not contain information on the signal amplitudes. This method does, however, make the comparison of the PSDs of the different signals easier by presenting a comparison of the dominant frequencies in the signals irrespective of signal amplitudes.

For correlation analysis, the data in a given session was divided into segments approximating sets of back and forth movements of the same speed and magnitude, deducted from the known motion pattern parameters. One experiment session of stage 1 comprised 18 segments and 1 session of stage 2 comprised 10 segments. A moving window cross-correlation (Matlab R2012a) was calculated for each of these segments, between the motion artifact, the impedance, and the programmed motion pattern. All the correlations were collected in respective matrices for stage 1 and stage 2. The medians of these matrices were calculated to give an overall correlation coefficient between the signals. The segmentation was done to account for a delay that was occasionally observed between the electrode–skin impedance and the surface potential. This delay was seemingly random and sometimes reversed, could be up to 200 ms, and was seen to be fairly steady when it existed for a given movement set.

The artifact-free ECG signal was calculated by subtracting the motion artifact signal, ch1 from the measured ECG containing also the motion artifact, ch2. From this noise-free signal, corresponding to the ECG measured between the chest location and the back of the hand, R-peaks were detected using a modified and simplified version of the Pan–Tompkins R-Peak detection algorithm (Pan and Tompkins 1985). The locations for the R-peaks detected from this derived signal were then used to locate the R-peaks on the original ECG signal, ch2, containing both the motion artifact and the ECG. This method bypassed the errors in R-peak detection from the original ECG that occurred due to motion artifact. The amplitudes of the R-peaks from this signal, for an experiment session, were calculated and their median values taken. To calculate a customized signal-to-noise ratio (SNR) to be used in comparing the effects of various parameter changes between sessions, this median value of R-peak amplitudes for a given session was divided by the root mean squared (RMS) magnitude of the motion artifact signal band-pass filtered between 0.4 Hz and 40 Hz for that session, as shown in equation (3). In this calculation, we assumed that because the main frequency components of the motion artifact generated in this study are much lower than that of frequencies contained in the QRS complex, the motion artifact would have a negligible effect on the median R-peak peak-to-peak value calculated over a multitude of peaks.

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\text{Custom SNR} = \frac{\text{median (R-wave amplitudes)}}{\text{RMS amplitude of motion artifact}}
\]

3. Results

Figure 5 depicts the results of a stage 1 session with increasing amplitude and speed (pattern 3). It shows that the electrode–skin impedance and the surface potential clearly follow motion and increase with an increased angle. The baseline skin–electrode impedance throughout the experiments was approximately 60 Ω. Figure 5 also shows the effect of the different low
cut-off frequencies on the low frequency drift. Figure 6 demonstrates that this drift is not caused by the system.

The effect of the magnitude of the electrode displacement, which was depicted in figure 4, on the motion artifact is also presented in figures 5, 7 and 8. In figure 7, the far left and the far right columns show this effect. The far left column shows the effect of increasing the rotation magnitude at a constant speed (pattern 1), and the far right column shows the effect of simultaneously increasing the extent of the rotation and the rotation speed (pattern 3). In figure 8, the last 25 s, the second half of each graph corresponds to a simultaneous change in speed and amplitude of the rotation of the servo, and thus also shows the effect of increased rotation range on the motion artifact.

The effect of movement frequency on the motion artifact is seen in the middle column of figure 7, where a stage 1, motion pattern 2 example is presented. The same effect is seen in the first 25 s of each graph in figure 8, which corresponds to the motion pattern 2 implementation in stage 2, increasing only the speed. The frequency of the main components of the motion artifact and the frequency of impedance change are determined by the frequency of the applied motion.

The PSD are shown in figure 9. In these graphs, it can be seen that the frequency components of the programmed servo motion, the motion artifact-related surface potential and the electrode–skin impedance are similar. To make the comparison easier, the PSD data is normalized to 1.
The cross-correlation between the impedance signal and the motion artifact signal, which were both band-pass filtered between 0.4 Hz and 40 Hz for all experiments in stage 1 using textile electrodes, is 0.56. The correlation between impedance and the programmed, thus known, movement pattern is 0.94. The correlation between the motion artifact signal and the movement pattern is 0.59.

The data from the stage 2 experiments using textile electrodes, band-pass filtered at 0.4 Hz to 40 Hz, produced correlations of 0.61 between the impedance and the motion artifact, 0.84 between the impedance and the movement pattern, and 0.68 between the motion artifact and the movement pattern. These correlations are shown in table 2.

The cross correlations of motion artifact, impedance, and the motion pattern for stage 1 and 2 sessions using the textile electrodes.

The custom SNR calculated for stage 1 data, as described in equation (3), are presented in table 3. A sample of the ECG signal together with the motion artifact, the electrode–skin impedance, and the programmed servo rotation is given in figure 10.

The averages of the SNR of all experiments for a given electrode at a given applied force for stage 1. SNR is defined as the average of the median R-peak peak-to-peak amplitude divided by the motion artifact RMS amplitude. The standard deviations are given in parentheses.

Figure 6. The output of the FlexiForce A201 sensor in the absence of electrode movement. Top left, 2 min of force reading of the calibration weight, after 5 min of calibration. Top right: 2 min of force reading while the electrode is manually pressed onto a kitchen scale. Centre left: 2 min of the electrode being pressed on the arm of a subject. Bottom left: the ECG (blue) and the electrode–skin impedance (green) during the same time. Centre and bottom right: a zoomed-in version of the centre and bottom left plots with 10 s of data presented. The units for the y-axes are the same horizontally, and x-axes units are the same for all.
The results of the SNR calculations for stage 2 are presented in table 4, and an example data set used in the calculations is shown in figure 11. The standard deviations of the SNRs are given in parentheses.

Pressing force has an effect on both the impedance and the artifact and how they are related to applied motion, as shown in figures 7 and 8. In figure 7, the upper row of data corresponds to the three motion patterns at 750 g of applied force. The lower row corresponds to the same procedure at 1 250 g of applied force. In figure 8, the rows are for a given setup of stage 2 and the applied force is increased from 250 g to 1 250 g in 250 g increments, from left to right. The same tendency for the effect of pressure is also seen in table 4 as an increase in the SNR with increasing pressure.

The motion artifacts and the impedance changes observed at the two locations resulting from the same movement under the same force are presented in figure 12. Differences in the resulting motion artifact for differing support structures between the textile electrode and the movement device can be observed. These differences are seen in figure 8, where the top row shows signals obtained using the first type of support, the 50 mm diameter foam padding, while the lower row shows signals obtained using the smaller 25 mm diameter foam padding. This effect is seen in table 4 where the SNR for the small foam padding is seen as consistently smaller than the SNR for the large foam padding.
In this paper, we introduced an accurate, controllable motion artifact generation system with integrated force measurement. The system creates the motion at the monitoring system—monitored user interface. This interface consists of the skin, the electrode layer in contact

**Figure 8.** The effects of applied force and different paddings. The data shown is from stage 2. The left row corresponds to the textile electrode with the 50 mm diameter foam padding. The right row corresponds to the textile electrode with the 25 mm foam padding. From top to bottom, the applied pressure increases from 250 g to 1250 g. Motion artifact is shown in blue and the impedance is shown in green. During the first half of each graph, only the speed of the motion is increased, while in the second half, the speed and the range of motion is increased simultaneously. The data has been band-pass filtered between 0.4 Hz and 40 Hz. To gain space for demonstration clarity, the names and values of the axes in the inner portions of the figure are not shown. Axes, units, and values are the same, left to right for the first y-axes, right to left for the second y-axes, and top to bottom for the x-axes.

### 4. Discussion

In this paper, we introduced an accurate, controllable motion artifact generation system with integrated force measurement. The system creates the motion at the monitoring system—monitored user interface. This interface consists of the skin, the electrode layer in contact
with the skin, and the structure between this electrode layer and the electrode mounting setup used to secure the electrode in place. Most motion artifact studies have been done so far by using user-initiated motions (Wiese et al 2005, Cho et al 2009, Pengjun et al 2011, Buxi

Figure 9. The normalized PSD’s of the impedance and the motion artifact and the motion pattern. The source signals are presented in the column on the right of figure 7 and have been band-pass filtered between 0.4 Hz and 40 Hz. Because of normalization, the PSD shown has no units. All y-axes have the same scale, and all the x-axes have the same scale.

Table 2. Cross-correlations.

|               | Motion artifact & impedance | Motion artifact & movement | Impedance & movement |
|---------------|-----------------------------|----------------------------|----------------------|
| Stage 1       | 0.56                        | 0.59                       | 0.94                 |
| Stage 2       | 0.61                        | 0.68                       | 0.84                 |

Table 2. The cross correlations of motion artifact, impedance, and the motion pattern for Stage 1 and Stage 2 sessions using the textile electrodes.

Table 3. Signal to Noise Ratios calculated for Stage 1.

| SNRs of Stage 1 | 750 g | 1250 g |
|-----------------|-------|--------|
| 50 mm Padding Distal | 42.7 (13.3) | 36.7 (18.8) |
| 50 mm Padding Proximal | 29.2 (10.1) | 30.6 (10.0) |
| Ambu Blu Distal | 149.3 (10.1) | 73.3 (19.0) |
| Ambu Blu Proximal | 79.3 (24.8) | 42.3 (32.3) |

Table 3. The averages of the signal to noise ratios (SNR) of all experiments for a given electrode at a given applied force for Stage 1. SNR is defined as the average of the median R-peak peak-to-peak amplitude divided by the motion artifact RMS amplitude. The standard deviations are given in parentheses.
The motion generally depends on user compliance with the testing protocols and those motions are prone to inter-user and inter-subject variations and are largely dependent on the design of the mounting system. Our system allows for motion artifact studies to be carried out under more controlled settings in a more reliable, thorough, and rapid manner. More importantly, this testing of the electrode–skin contact of the monitoring system without the effect of the garment’s electrode mounting setup makes this an important precursor to whole system tests. By isolating the motion to the electrode and the contact point, this system has the potential to aid the design of an electrode more resilient to motion, for any given mounting system. The ability to analyse the effects of applied force on the electrode may also be used to set design guidelines for the mount system.

The electrode to be studied is pressed onto the designated anatomical location in a force-controlled manner. This electrode is moved by a servo that is controlled by a microcontroller, which is programmed through a PC interface. As the motion is generated, the ECG, the motion artifact and the impedance and a depiction of the servo rotation of a stage 1 session. An electrode with 50 mm foam padding is located on the proximal forearm location, and pressed with a 750 g force. The data shown corresponds to pattern 1 motion change, where the motion amplitude is increased and band-pass filtered between 0.4 Hz and 40 Hz.

Table 4. Signal to Noise Ratios calculated for Stage 2.

|                  | 250 g | 500 g | 750 g | 1000 g | 1250 g |
|------------------|-------|-------|-------|--------|--------|
| 50 mm Padding    | 8.0 (4.0) | 16.3 (9.6) | 18.7 (15.3) | 25.3 (18.4) | 21.0 (13.4) |
| 25 mm Padding    | 7.0 (6.7) | 9.2 (5.3)   | 7.7 (4.4)    | 10.0 (4.0)  | 11.3 (4.5)  |

Table 4. The SNRs for Stage 2. The standard deviations of the SNRs are given in parentheses.

et al 2012). The motion generally depends on user compliance with the testing protocols and those motions are prone to inter-user and inter-subject variations and are largely dependent on the design of the mounting system. Our system allows for motion artifact studies to be carried out under more controlled settings in a more reliable, thorough, and rapid manner. More importantly, this testing of the electrode–skin contact of the monitoring system without the effect of the garment’s electrode mounting setup makes this an important precursor to whole system tests. By isolating the motion to the electrode and the contact point, this system has the potential to aid the design of an electrode more resilient to motion, for any given mounting system. The ability to analyse the effects of applied force on the electrode may also be used to set design guidelines for the mount system.

The electrode to be studied is pressed onto the designated anatomical location in a force-controlled manner. This electrode is moved by a servo that is controlled by a microcontroller, which is programmed through a PC interface. As the motion is generated, the ECG, the motion artifact and the electrode–skin impedance are simultaneously measured. Previously, only Ödman et al and Liu et al (Ödman and Öberg 1982, Liu et al 2013), have developed somewhat similar systems. Ödman made a motion generation system for gelled surface electrodes, without the monitoring of applied force. Liu et al designed the system for testing dry electrodes.
coupled with electrolyte gel on a membrane acting as artificial skin. According to Ödman, the deformation of the gel and the motion of the electrode relative to the gel produced smaller potential variations when compared to the potentials created by the movement of the electrode or the gel on skin and the potentials created by skin deformation. Thus, a system that creates a motion artifact on the skin could be used to improve electrode design relative to skin contact and movement transmission from the mounting system to the skin, before the real world application of the system is tested.

Our results confirmed that the designed motion artifact generation and assessment system functions as intended and indicate that the motion artifact and the electrode–skin impedance clearly follow the applied motion. The system generated the controlled motion and this in
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Figure 12. The different locations for the same padding with the same force. The first row is the electrode with the 50 mm foam padding, located on the distal forearm. The second row is the same electrode located on the proximal forearm location. Shown are two motion patterns: motion change pattern 2 (left) and motion change pattern 3 (right). As an example, a difference in the rising edge can be seen in the left graphs. A 1 s sample of the ECG with halved amplitude is presented in red, for comparison. The data has been band-pass filtered between 0.4 Hz and 40 Hz. Axes, units, and values are the same left to right for the first y-axes, right to left for the second y-axes, and top to bottom for the x-axes.

The programmable motion allows for two key things. First, it allows the repetition of motion to be kept standard; having a known cause for the possible changes observed in the measured signals. Secondly, the motion speed, frequency and range can be programmed and altered to get a full range of measurements in the function of frequency, range or applied force. The ability to also change the motion pattern to non-linearly changing motion enables the better fit of the motion to the study or application-specific motion characteristics. This ability can also be used to study the elastic properties of the skin related to the motion artifact, as various movement frequencies and displacement magnitudes might be necessary to study the nonlinear viscoelastic properties of the skin.

The results indicate that the amplitude of the motion artifact is related to the electrode displacement magnitude in a non-linear relationship. The maximum electrode displacement
range that can be examined is limited by the elasticity of the skin. After this point is reached, the electrode will move on the skin, slipping and changing its location relative to the outermost layer of skin, and the experiment will fail in its objective.

It can be observed in both the time domain and the frequency domain that the dominant frequency of the motion artifact and the dominant frequency of the impedance variation are directly related to the frequency of the electrode movement. In the time domain, this observation for the motion artifact is done by visual inspection of the signals, as the correlation between the motion artifact to the applied motion is less than the correlation between impedance change and the applied motion. If for a given application the movement of the subjects is known, the motion artifact study can be narrowed down to the movement frequencies or even the movement patterns in question. The ability to program frequency can be used to tune the experiment to specific applications using frequencies that are the most problematic for the said application.

We also observed a low frequency drift happening at around 0.2–0.3 Hz. This drift was filtered away in our results as it was not caused by the applied motion. We think that the drift was due to the changes in the skin brought upon by the changes in blood flow with every heartbeat, or the movement of the thorax with breathing affecting the shoulder and thus the lower arm by proxy. The effect that this motion has on the applied force levels stays within the boundaries of the ±100 g tolerance for the applied force for a given force level.

The difference in the cross-correlation values between the motion artifact signal, impedance change and the programmed movement arises from the motion artifact signal containing components at different frequencies than the applied motion. The components with lower frequencies exist under 1 Hz, the higher frequency components are up to 7 Hz. Similar components are absent or of negligible amplitude in the impedance signal. These higher frequency components, even if they are very small compared to the main component of the motion artifact components at the movement frequency, cause a reduction in the correlations due to the nature of the cross-correlation function. This is because any deviation between the compared signals lowers the correlation, no matter how small the deviations are relative to the dominant waveforms of the signals. Another factor in the lower correlations could be the slightly different slopes of the signals, the small delays between the impedance change and the potential change, in response to motion, and the differing shapes of the peaks of these signals at instances where motion changes direction.

The differences between the shape and behaviour of the electrode–skin interface and the surface potential had been observed by other researchers as well. Ödman demonstrated that the impedance and surface potential have a different shape and state that the relationship between these two signals is a non-linear one (Ödman 1981). Later, Ödman and Åke Öberg stated that the highest potentials originated from the skin, and considerable potentials were generated at the electrode gel–skin interface, and that these were not dependent on the impedance change (Ödman and Öberg 1982). Three theories on the origins of skin potentials are presented in a later paper by Vos et al (2003). Ottenbacher et al (2008a) mentioned the assumption that the impedance change arises mostly at the changing area of the electrode contact rather than the skin itself. Their results also show this difference in the behaviour between impedance and potential. From these, we think that the difference arises from a combination of factors: besides the coupling fluid, it could also be that the viscoelastic properties of the pathways causing the impedance change and the potential change might be slightly different, especially if they partly originate from different layers of skin, albeit in the same millimetre thick skin segment.

As our results clearly demonstrate, controlling the force is necessary for repeatable and comparable results. The motion artifact and the impedance relationship to the applied motion
depend on the applied force. As most dry electrodes make use of tight garments to keep them in place, knowing this applied force will be useful in the garment specifications related to elasticity and tightness. The measurement of the applied force is, therefore, essential in order to normalize the environment for cross-relating separate studies and also for aiding or modelling the design of the wearable garment and electrode. We would like to note that, for an electrode in a garment, the force on the electrode will not only change because of the changes in skin and tissue under the electrode, but also because the body movement will change the shape of the garment, changing the pressure applied by the garment to the body and the force applied to the electrode by the garment. So we decided to restrict our study to monitoring the initial condition of applied force, and focus on the movement.

In the present system, the force application is manual even if it is constantly monitored. As shown above, the force remains stable. For future systems, the electrode-mounting task can be motorized and controlled by a feedback loop involving the force sensor. It should be noted that the force sensor measures force applied by the mounting system to the electrode, which corresponds to the force applied by the garment to the electrode, not the pressure applied by the electrode to the skin. Of course, the force–pressure relationship with regard to applied force in relation to the electrode area can be deducted to monitor the pressure applied by the electrode, if that is more useful for a given application. The calibration, at present done manually, can be automated at the microcontroller level to speed up the operation.

The stage 1 experiments carried out at two different locations showed that the motion artifact originating from similar motions is different for each anatomical location. This is due to the different skin and tissue properties at the electrode location. Thus, the intended usage location should be considered when selecting the proper location to test the electrodes. This also indicates that, most probably, the difference in skin and underlying tissues in different individuals, for example, fat and other tissue types, do have an effect. Similar differences in the motion artifact might arise even with small position changes at the same location. Thus, this system can also be used to find the optimal positioning for electrodes for a standardized garment.

The testing of the two different structures showed that it is likely that some structures consistently behave better than others. For example, from the stage 2 data, it can be clearly observed that when bigger support padding is used between the mounting system and the electrode there is a considerable reduction in the motion artifact, when compared to using padding that is the same size as the electrode. Because these results are repeatable, the system is suitable for the testing of different electrode designs in order to optimize functional design parameters to further improve the stability of the electrode.

As this system enables the motion artifact-related tests to be carried out by separating the various factors that cause skin stretch, which is the main source of the motion artifact, the system enables detailed, thorough testing to be carried out in a rapid and repeatable manner. Monitoring the applied force and having the same motion parameters makes these studies comparable and useful at an earlier stage in the mobile monitoring system design and testing process. The resulting improvement in the electrode design, independent of the effects of the mount design, will translate into improved signal quality for the integrated garment.

The system prototype has proven functional, and there are numerous possibilities for improving the design. Even if currently stable, the manual operation of the lever to mount the electrode to its intended location can be automated, saving operator time and further reducing any possible irregularity in the applied force. The movement which is programmed to be a linearly propagating back and forth rotation of the servo axis can be programmed to be a sinusoidally propagating back and forth rotation, providing clearer frequency peaks for analysis. An environment temperature sensor and skin humidity and temperature sensor can
be implemented for a better understanding of possibly relevant factors. As the skin is not a uniformly flat surface, torque sensors and multi-axis force sensors and accelerometers for the parallel existing forces and motion could increase the understanding of the events happening at the skin–electrode interface. A hi-speed camera could visually record skin deformation around the electrode.

5. Conclusion

A system that generates controllable and repeatable electrode movement during vertically applied steady and monitored force and simultaneously measures the ECG, motion artifact, and electrode–skin impedance, in order to study the electrode motion artifact, has been developed. In this study, we show that the system works as it was intended and that the affordable design has been proven to be functional. It can be used to assess the motion artifact in a thorough and repeatable manner, while focusing on the relationship between electrode movement and the resulting motion artifact. The system is constructed from commercially available parts and is partially 3D-printed. The designed system generates the reliable and repeatable motion of surface electrodes for motion artifact studies. The system helps to overcome the potential problems associated with the human factor in creating motion artifacts and the potential effects of the design of the mounting setup used in most of the previous studies.

The versatility of the system is achieved by making motion programmable and by implementing mount force monitoring. The speed and the amplitude of the motion can be adjusted for each consecutive motion step. As well as controlling the frequency of the motion, this adjustment makes both linear and non-linear patterns of motion possible.

Furthermore, as it is shown to be an important factor that affects the relationship between applied motion and the resulting motion artifact and impedance changes, force monitoring makes the repeatable and comparable measurement under constant force possible.

The developed system prototype has proven functional in the generation and assessment of motion artifacts. It can be used in experimental and developmental stages preceding the advanced motion artifact-related testing of the integrated garment, during which the garment will directly apply movement-causing forces to the electrode. This system is especially suited for thoroughly investigating the extent to which this motion artifact is influenced by various electrode designs. The system does not require the use of complex experiment protocols that need to be correctly taught to the subject or operator and closely followed. Reducing the electrode’s susceptibility to cause motion artifacts as a result of movement applied to it has the potential to increase the biosignal monitoring system’s tolerance to movement, independent of the electrode mounting setup design.

In addition to these practical tests that are geared towards the design and production of electrodes and to help in the design of wearable systems for a wide scope of applications, the system can also be used for the study of basic phenomena of motion artifacts to help study motion artifact origins and to test various skin and tissue types. Therefore, the system can be used in studies concerning the theoretical concepts related to the motion artifact.

Further automation of the various aspects and the addition of sensors will make the system more reliable and easier to use, and increase the scope of obtained data.

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