HEART RATE DETECTION FROM IMPEDANCE PLETHYSMOGRAPHY BASED ON CONCEALED CAPACITIVE ELECTRODES

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Abstract − Physiological monitoring outside clinical environments and medical offices can contribute to people's wellness, particularly if that monitoring does not disturb their daily activities, and can also help in life-style improvement. We propose a novel non-conscious method to monitor the heart rate which is based on detecting blood volume changes in the thigh through electrical impedance plethysmography while the subject is seated. Four conductive strip electrodes are concealed underneath seat's upholstery, hence working as capacitive electrodes. A 50 kHz current is injected trough two outer electrodes and the changes in the drop in voltage between the inner electrode pair are detected.

Tests performed with a system designed in our laboratory show that the recorded signal has peaks that unmistakably coincide with the heart rate. The waveform depends on the relative position of both pairs of electrodes, and the sensitivity to impedance changes depends on the position and area of the voltage-detecting electrodes with respect to those of the current injection electrodes. The amplitude and waveform of the recorded signal are good enough for the heart rate to be straightforwardly obtained by a slope-detection-based algorithm.

Keywords: heart rate detection, impedance plethysmography, non-conscious monitoring

1. INTRODUCTION

Daily health monitoring outside clinical environments or medical offices is instrumental to detect possible health disorders before they become important [1] and for life-style improvement, which can help in preventing illnesses. It can also help in managing people with chronic diseases and for elder's care, thus contributing to their wellness.

Home health monitoring offers an increased comfort for patients, which is especially important for older people, patients with chronic diseases and people with disabled mobility, as it can be difficult and expensive for them to visit a hospital or medical office for routine checks. For the best advantage, health monitoring during daily life should be performed with a minimal intervention of the subject and without disturbing his or her daily activities [2]. Non-conscious measurements fulfil these requirements. One approach is to measure without any contact with the subject.

Attempts to record biomedical signals without any physical contact can be traced at least back to the 1970's when Lin *et al.* [3] recorded the apexcardiogram by sending microwave radiation to the chest and detected the echo signal modulated by the precordial movements.

More recently, Harland *et al.* [4] developed a sensor-amplifier with an extremely high input impedance to detect bioelectric signals without any physical contact with the body. They recorded the heartbeat with an electrode placed at 1 m from the chest, as well as the electroencephalogram (EEG) [5] with an air-gap of 3 mm between the hair and the electrodes. The actual signal detected when measuring voltage from a distance from the body surface without any contact is still in dispute as distance variations between body and electrodes can contribute to the voltage detected [6]. Harland *et al.* also obtained the electrocardiogram (ECG) from a few millimetres from the wrist with capacitive electrodes (which implies a physical but non-conductive contact) using a wristwatch-type interface [7]. Therefore, this is a wearable technology, which results in a minimal disturbance but cannot be properly considered non-conscious.

Measurements that rely on a direct contact between bare skin and sensor do not necessarily disturb daily activities either. Tanaka *et al.* [8] measured the blood pressure (BP) for a person seated in a toilet. They used the volume-oscillometric method with a detection (contact) area of 40 mm in the thigh. A servo motor applied the force and the blood volume changes were detected with an optical sensor. The same authors proposed a method to compensate for the hydrostatic pressure to calculate BP at heart level. Ishijima [1] recorded the ECG with electrodes placed on each side of the toilet seat. When a person seats down, their thighs come into direct contact with the electrodes.

The measurements above, however, last for a short time only, so that they are periodic rather than continuous. Non-conscious measurements using concealed sensors in furniture can provide longer periodic records or even continuous records if motion artefacts are reduced. Using non-conductive electrodes covered with the upholstery of a chair, Lim *et al.* [9] measured the heart rate (HR) through the record of the ECG for a seated person wearing normal clothes. The amplitude of the detected ECG heavily depended on the proximity of the body to the electrodes as their capacitance decreased when separating the body from the back of the seat, whereas the amplitude of the surface

ECG obviously remained the same. Measurements involving an excitation signal can compensate for an increased separation between the body and the electrodes by using a stronger excitation.

Based on this concept, we propose to measure the heart rate by capacitive impedance plethysmography (IPG) when the subject is seated. By applying a large-enough excitation signal, electrodes can detect impedance changes in the thigh through clothes and upholstery. The subject is unaware of the measurement being performed.

2. MEASUREMENT METHOD

Impedance plethysmography is the measurement of volume changes through the measurement of the electrical impedance of the body tissues of interest [10]. When measuring in limbs, these volume changes are mainly due to the blood flow. Common IPG measurements in a limb use four band electrodes around it [11] to obtain a homogeneous electric field in the target volume. Schraibman *et al.* [12] obtained the IPG by attaching to the calf disposable spot electrodes instead of the usual circumferential electrodes. The signal detected was more affected by electrode placement, which may be important to estimate blood flow, but it is not essential for heart rate estimation.

We propose to use a tetrapolar arrangement of flat electrodes (Fig. 1) placed under the thigh. The outer electrodes are used to inject a current and the inner electrodes measure the drop in voltage across the tissues above them. If the electrical conductivity of those tissues changes, for example because of blood flow, the voltage detected will have the same frequency as the applied current and its amplitude will depend on the basal impedance and on those conductivity changes. Our electrodes are built from conductive adhesive tape that is thin enough to get unnoticed when placed underneath seat's upholstery. Because excitation and detection are carried out through different electrode pairs, electrode-tissue impedance will have no effect, provided the amplifier has a high-enough input impedance [13].

2.1. Electrode's size ad position

The distance D (Fig. 1) between the current injection electrodes should be as large as feasible, so as to cover as much area as possible of the leg segment being measured. The optimal separation d between voltage-detection electrodes, however, is not known in advance.

Some authors have studied the sensitivity to local conductivity changes of bipolar and tetrapolar transfer impedance measurements in biological tissues using either circumferential electrodes around a uniform anisotropic cylinder [11, 14] or point electrodes in a semi-infinite homogeneous medium [15, 16]. These studies have shown that tetrapolar measurements are less sensitive than bipolar measurements but are more robust to artefacts from superficial layers close to the electrodes. Further, the sensitivity for bipolar measurements decreases with the tissue depth but for tetrapolar measurements in a semi-infinite homogeneous medium it is maximal at a depth that depends on the ratio between the spacing between current

injection electrodes and that between voltage detection electrodes.

For circumferential electrodes, the closer the two pairs are, the higher the sensitivity to vessels near the surface is. If the distance between the current injection pair increases and the distance between the voltage detection pair remains the same, the sensitivity to deeper vessels relative to surface vessels increases, but not linearly [11]. Further, the relationship between electrode's separation and sensitivity to deep vessels depends on the limb's diameter. Our tetrapolar array of strip electrodes, however, must fit any seated adult, and therefore it must be designed so that current is always injected into the thigh regardless of its dimensions.
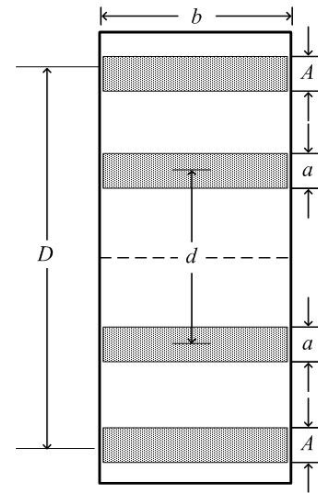

Fig. 1. Tetrapolar array of strip electrodes to measure the IPG in the thigh. Both pairs of electrodes are symmetrically placed with respect to the centre line of the array (dashed line).

Point electrodes can replace circumferential electrodes in blood flow measurements based on impedance plethysmography if they are in direct contact with bare skin of the calf [12]. However, the best results are obtained when the electrodes are in opposite faces of the limb (current injection electrodes in the anterior face and voltage detection electrodes in the posterior face.) For a concealed measurement, the electrodes must be placed under the thigh. Besides, electrodes in our linear array are on a flat surface whereas pairs of spot electrodes can define different axis even if placed on the same face of the limb, which may be advantageous to target specific voxels. On the other hand, the area of spot electrodes is fixed to that of commercially available models, but in our linear array we can design different electrode areas, in addition to selecting the distance D between current injection electrodes and the spacing d between voltage detection electrodes. With regard to the length b of the electrodes, it should be longer than the diameter of the thigh for a seated person, so that the measurement be as independent as possible of the position of the thigh with respect to the chair. The width A of current injection electrodes must be large enough to obtain a uniform distribution of current lines across the thigh. The effect of the width a of the voltage detection electrodes on the detected signal is a variable to be analyzed.
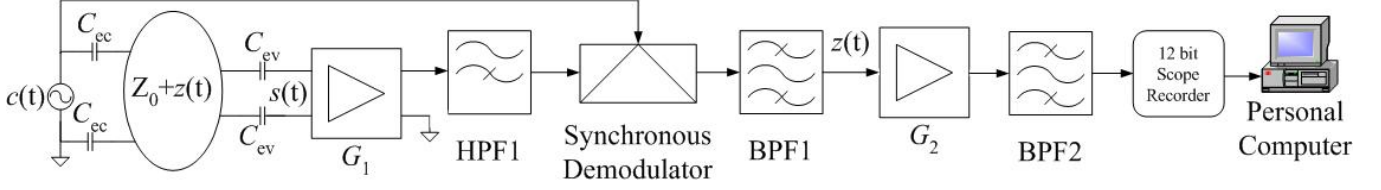
Fig. 2. IPG measurement system. Impedance changes due to blood flow in a body segment are synchronously demodulated by using the carrier signal, c(t), as a reference. The time-invariant component of bioimpedance is rejected by BPF1. G_2 amplifies the small changes of IPG and BPF2 eliminates the respiration component and reduces the noise bandwidth.

2.2. Signal detection

Fig. 2 shows the proposed IPG measurement system. We inject a 50 kHz sine wave carrier c(t), which is high-enough to reduce the impedance of capacitive electrodes C_{ec} and C_{ev}, yet not that high as to require amplifiers with a very large gain-bandwidth product. The amplitude-modulated signal s(t) measured by the detection electrodes C_{ev} is the product of c(t) times the basal impedance Z_0 plus the blood-flow-related signal z(t). As a result, the frequency of this signal f_z is upward translated to yield two side bands around the carrier frequency f_c. We recover $Z_0 + z$(t) by coherent demodulation using the excitation signal as a reference. Before demodulating, the differential voltage detected is amplified by an instrumentation amplifier (IA). Because the equivalent signal source is capacitive, the front end includes a resistor-T network (R_1, R'_1 and R_2 in Fig. 3) to provide a path to ground for input bias currents. These resistors can be selected large enough for the impedance from each signal line to ground to be determined by the input capacitance of the corresponding input of the amplifier (C_i, C'_i). Therefore, each sampled voltage will be attenuated according to

$$\frac{V_i}{V_z} = \frac{C_{ev}}{C_{ev} + C_i} \qquad (1)$$

Once amplified by G_1, the signal is high-pass filtered to block the offset voltage (HPF1).
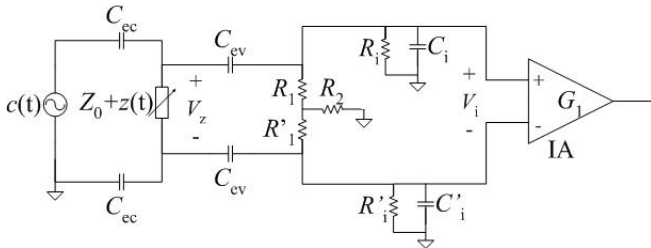

Fig. 3. Input equivalent circuit for IPG measurements. Input capacitive electrodes C_{ev} form a voltage divider with the input capacitance C_i and $C_i^{'}$ of the IA.

The synchronous amplitude demodulator can be implemented by a switched-gain amplifier, controlled by a symmetrical square waveform obtained from the carrier signal through a voltage comparator [17]. After demodulation, a band-pass filter (BPF1) rejects both the dc components (offset voltage and basal impedance) and any harmonics that result from the demodulation. Because

impedance variations produced by blood circulation are in the range from 0,1 % to 1 % of the total impedance [18], the signal is further amplified by a single-ended amplifier with gain G_2 before being band-pass filtered again to attenuate respiratory components and to reduce noise bandwidth (BPF2).

3. EXPERIMENTAL RESULTS AND DISCUSSION

As a first step to determine the position and size of the electrodes to obtain the best results, we performed a series of experiments for different d/D ratios and areas of the voltage-detection electrodes. First, we varied the distance d for a fixed width a for visible electrodes placed on the seat, so that the only separation between electrodes and leg were the subject's trousers. Then we reduced a and changed d/D again. Finally, we recorded the IPG when electrodes were covered by synthetic fabric upholstery. Table 1 summarizes the experiments. The fixed dimensions were $D = 27$ cm, $b = 22$ cm and $A = 5$ cm. D and b were selected so that the array was larger than the area covered by the thigh of a seated person.

The supply voltage rails were ±12 V, the peak-to-peak voltage applied was 10 V, and G_1 was adjusted for each subject and experiment to obtain the maximal output in the IA. This ensured that the amplitude of s(t) after G_1 was the same for all experiments. The IA (INA111) has: common mode input impedance $R_i = 10^{12}$ Ω shunted by $C_i = 3$ pF and input bias currents 2 pA (typ.). We selected $R_1 = R'_1 = 1$ MΩ and $R_2 = 10$ MΩ. At 50 kHz the impedance of C_i is 1 MΩ, hence much smaller than the common mode input resistance. If R_i were too small, C_{ev} and R_i would result in a high-pass filter that could attenuate the modulated signal.

The corner frequency of the filters in Fig. 2 were $f_{L1} = 1,6$ kHz for HPF1, $f_{L2} = 0,5$ Hz and $f_{H2} = 10$ Hz for BPF1, and $f_{L3} = 0,5$ Hz and $f_{H3} = 10$ Hz for BPF2. All filters were first-order passive filters. The demodulated signal was amplified by $G_2 = 3000$.

Table 1. Experiments to the measurement of capacitive IPG. $G_2 = 3000$ in all experiments.

Experiment	$d/D =$	$a/A =$	$G_1 =$
1	0,2	1	16
2	0,6	1	6,7
3	0,2	1/4	34
4	0,6	1/4	7,5
5 (concealed)	0,2	1/4	500

The photoplethysmogram (PPG) in a finger was recorded simultaneously with the IPG to obtain a reference signal. Each signal was digitized at 2000 samples each second by a 12 bit scope recorder (Yokogawa). A PC processed the digitized signals. All measurements shown here were obtained in the same leg of the same person.

Fig. 4 shows the IPG (top) and the PPG (bottom) signals obtained from one volunteer. The IPG was obtained with voltage detection electrodes placed at a distance $d = 0.2 \times D$ apart and width $a = A$ (experiment 1). An upward peak in the IPG, which means a decrease in impedance, is clearly visible at each cardiac cycle.

Next we increased the separation between voltage-detection electrodes to $d = 0.6 \times D$ while keeping $a = A$ (experiment 2). Fig. 5 shows that the IPG does not display a clear peak at each heart beat, which means a reduced sensitivity to blood flow.
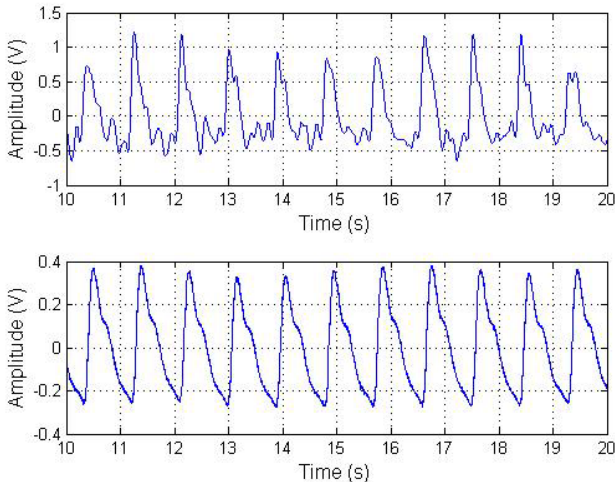

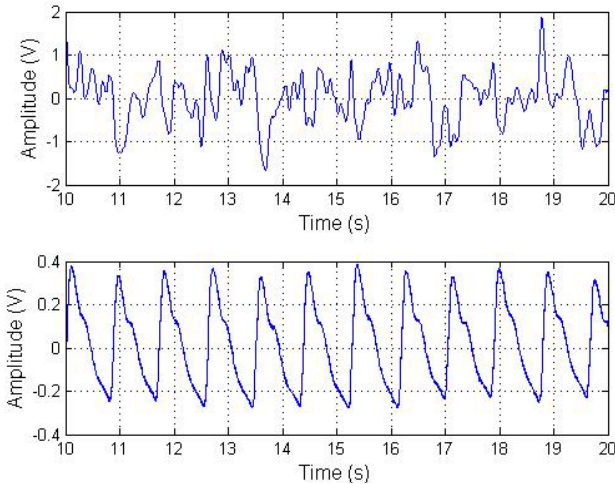Fig. 4. Experiment 1. IPG (top) obtained with $d/D = 0.2$ and $a/A = 1$, and PPG (bottom).

Fig. 5. Experiment 2. IPG (top) obtained with $d/D = 0.6$ and $a/A = 1$, and PPG (bottom). The heart rate is not distinguishable to the naked eye from the IPG.

To evaluate the effect of the width of the voltage detection electrodes, we reduced it to a = A/4 and set the distance back to $d = 0.2 \times D$, as in experiment 1. The result for this experiment (experiment 3) is shown in Fig 6.

Upward peaks synchronous with each heart beat are clearly present in the IPG and their amplitude is about eight times larger than that obtained in Fig. 4 for wider electrodes. To assess to what extent this was due to an increased sensitivity for narrower electrodes we increased the distance d to $0.6 \times D$ (experiment 4). Fig. 7 shows the recorded signals. The amplitude of the IPG decreased by 5 but was still larger than that for wider closer electrodes (Fig. 4) and the heart rate was clearer than the measurements with wider electrodes placed at the same distance (Fig. 5).
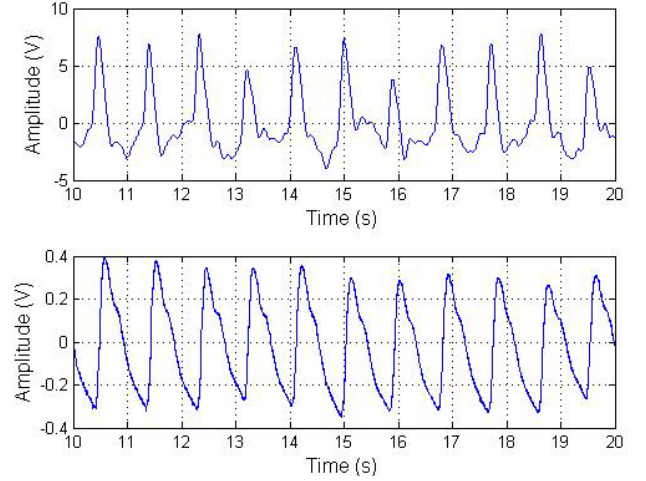

Fig. 6. Experiment 3. IPG (top) obtained with $d/D = 0.2$ and $a/A = 1/4$, and PPG (bottom). The IPG is larger than that in Fig. 4, where d/D was the same but a/A was larger.
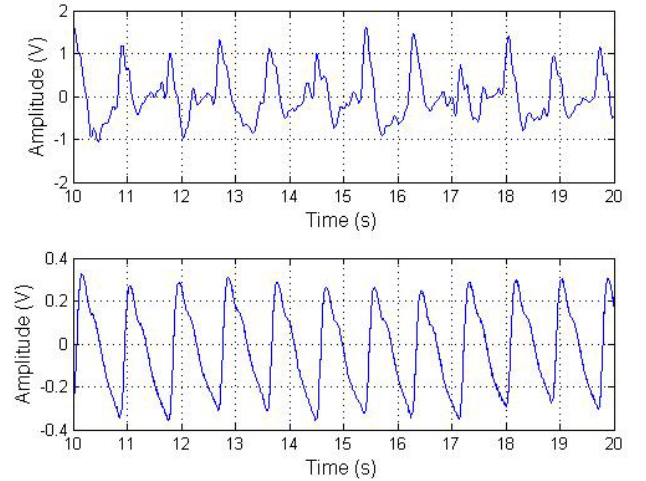

Fig. 7. Experiment 4. IPG (top), measured with $d/D = 0.6$ and $a/A = 1/4$, and PPG (bottom). Despite of the IPG is smaller than that in Fig. 6, obtained with the same a/A but smaller d/D ratio, the heart rate is clearly distinguishable. This was not true for wider electrodes placed at the same distance (Fig. 5).

By comparing the IPG in Figs. 4 and 6, we also realize that in addition to the larger IPG in Fig. 6, peak fluctuations because of the respiration also seem larger when using narrower electrodes (Fig. 6). Fig. 8 shows the spectrum of

both signals and confirms that narrower voltage detection electrodes better detect respiration-related impedance changes.

For circumferential and spot electrodes, it has been reported that the sensitivity to deeper blood vessels reduces when the d/D ratio increases. Here we have observed the same effect for our linear array (Figs. 4 and 5). However, for any d/D ratio, the amplitude of the IPG can be increased by reducing the width of the voltage detection electrodes (Figs. 6 and 7).
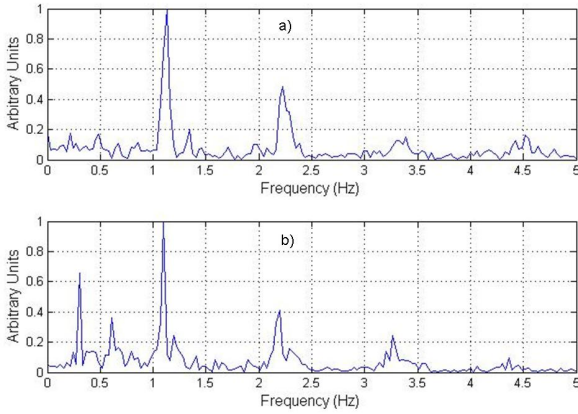

Fig. 8. Power spectral density of the IPG obtained with $d/D = 0,2$ and a) $a/A = 1$, and b) $a/A = 1/4$. Low frequency components (attributable to the respiration) are larger for the narrower electrodes.

Since the larger IPG was obtained with the narrower electrodes placed at a short distance d apart, we selected $a = A/4$ and $d = 0,2 \times D$ to record the IPG with concealed electrodes (experiment 5). Fig. 9 shows that the amplitude of the IPG decreased by about 50 % as compared to that obtained with visible electrodes (Fig. 6) and was noisier. Fig. 10b confirms that the SNR decreased to the extent that the spectral contents at the respiratory frequency is almost buried in noise whereas in Fig. 10a, obtained with visible electrodes, frequency bands at the respiratory frequency were obvious. Nevertheless, the amplitude and waveform of the IPG obtained with concealed electrodes were still good enough for the heart rate to be easily extracted by identifying the peak in each cardiac cycle with a slope-detection-based algorithm. The vertical lines in Fig. 11 show the peaks of the IPG located with this algorithm.

Table 1 show that Z_0 was larger (smaller G_1) for separated voltage detection electrodes. Further, by increasing the distance between the electrodes and the leg (for example, by concealing the electrodes), a much larger G_1 is needed. Even so, the heart rate is clearly distinguishable.

The IPG recorded with flat capacitive electrodes is susceptible to movement artefacts, and these will be present in prolonged records. However, if the main purpose is to monitor the state of the subject, movement implies activity. If there were no activity, there would no movement and the IPG would reveal whether the heart is beating or not. Further, records during resting periods could serve for periodic monitoring. Knowing that even during a nap, one's

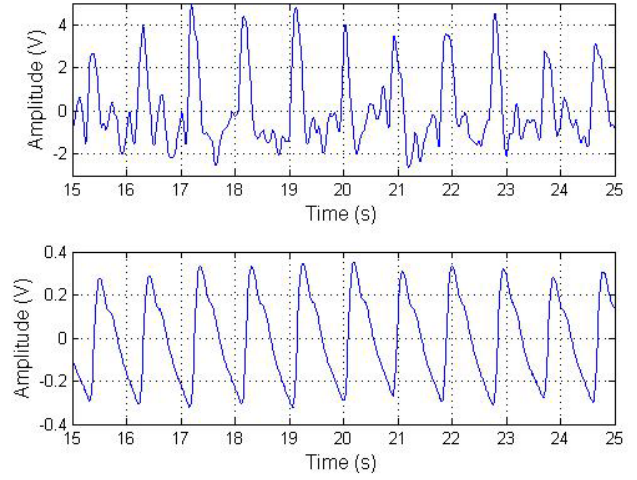health is being monitored helps in feeling confident and safe.

Fig. 9. Experiment 5. IPG (top), measured with concealed electrodes with $d/D = 0,2$ and $a/A = 1/4$, and PPG (bottom).
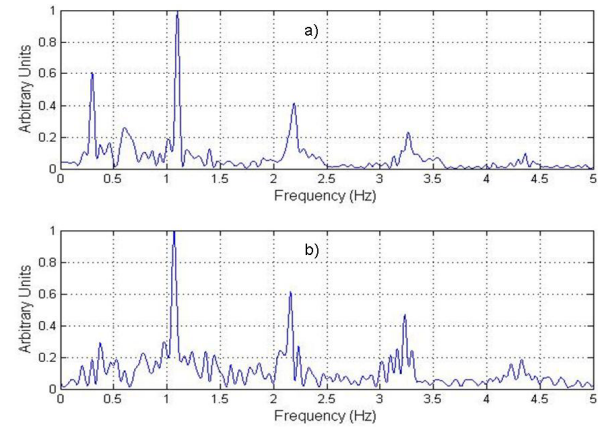

Fig. 10. Spectral content of IPG measured with a) visible and b) concealed electrodes. Respiration-related information is clearer with visible electrodes.
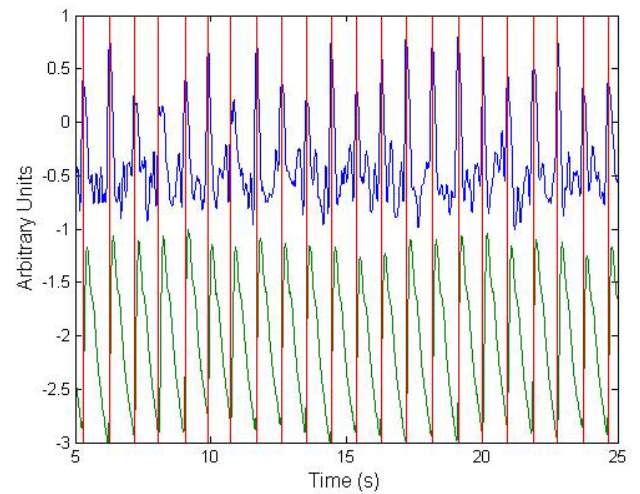

Fig. 11. Heart beats (vertical lines) detected from the IPG (top signal) by a slope-detection-based algorithm.

4. CONCLUSIONS

A novel non-conscious method to measure the heart rate has been described which is based on detecting blood flow in the thigh through impedance plethysmography obtained with a tetrapolar array of strip capacitive electrodes placed under the thigh (Fig. 1). This approach can be thought of as impedance plethysmography based on circumferential electrodes that had been "opened." Therefore, qualitative conclusions regarding the relationship between electrode separation and sensitivity to deep vessels in conventional impedance plethysmography offer a first clue to design an electrode array concealed beneath a seat's upholstery.

We have performed five experiments to find the qualitative relationship between the distance and width of the voltage detection electrodes and the waveform of the IPG. For the distance, the smaller is the d/D ratio, the larger is the IPG (Fig. 4 compared to Fig. 5, and Fig. 6 compared to Fig. 7). This agrees with results in the bibliography for spot and circumferential band electrodes. For any given d/D ratio, the IPG increases when the width of the voltage-detection electrodes decreases (Fig. 4 compared to Fig. 6 and Fig. 5 compared to Fig. 7). The amplitude of the IPG decreases when the distance between electrodes and thigh increases, for example, when electrodes are beneath the upholstery (Fig. 9 compared to Fig. 6). Further work is needed to find the optimal values for the d/D and a/A ratios, where D is determined by the depth of a seat, with regard to both the amplitude of the IPG and the detection of impedance variations related to the respiration.

Since the electrodes can be concealed underneath the upholstery of a normal chair, sofa or cushion, the method proposed can provide a useful mean for continuous heart rate monitoring in non-clinical settings. In the absence of movement artefacts, the heart rate can be extracted from the IPG by a simple slope-detection-based algorithm.

ACKNOWLEDGMENTS

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