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Development of a smart backboard system for real-time feedback during CPR chest compression on a soft back support surface

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***Abstract*—The quality of cardiopulmonary resuscitation (CPR) is often inconsistent and frequently fails to meet recommended guidelines. One promising approach to address this problem is for clinicians to use an active feedback device during CPR. However, one major deficiency of existing feedback systems is that they fail to account for the displacement of the back support surface during chest compression (CC), which can be important when CPR is performed on a soft surface. In this study we present the development of a real-time CPR feedback system based on an algorithm which uses force and dual-accelerometer measurements to provide accurate estimation of the CC depth on a soft surface, without assuming full chest decompression. Based on adult CPR manikin tests it was found that the accuracy of the estimated CC depth for a dual accelerometer feedback system is significantly better (7.3% vs. 24.4%) than for a single accelerometer system on soft back support surfaces, in the absence or presence of a backboard. In conclusion, the algorithm used was found to be suitable for a real-time, dual accelerometer CPR feedback application since it yielded reasonable accuracy in terms of CC depth estimation, even when used on a soft back support surface.**

I. Introduction

Many previous studies have shown that, even when performed by trained clinicians in a hospital setting, the quality of cardiopulmonary resuscitation (CPR) is often inconsistent and frequently fails to meet published guidelines [1-2]. This can be attributed in part to the poor quality (i.e., depth, rate and delivery time) of chest compression (CC) during CPR [3-4].

One approach to improve the quality of CC during CPR involves using a feedback device to help clinicians deliver CC at the European Resuscitation Council (ERC) recommended rate and depth. Among the many challenges involved in developing such a device is the need to prevent information overload to the rescuer, since this has been shown in previous studies to be detrimental to CC performance during CPR [5-6]. For instance it has been reported that overemphasis on the CC rate may lead to the delivery of inappropriate net CC depths [7-8]. In addition, while the CC rate can be efficiently guided by a metronome, CC depth must be measured and processed to provide feedback. In the past this has been accomplished by measuring the sternal force applied using devices such as the CPR-Ezy. However, accurate estimation of the CC depth from force measurements depends heavily on the patient’s thoracic stiffness and damping [9-11]. Other feedback devices, such as pocketCPR® rely on an accelerometer to deduce the sternal displacement during CC. However, on soft surfaces such as a hospital mattress this approach is unreliable since the CC depth is overestimated due to the compression of the back support surface. The accuracy of the CC depth measurement can be improved by using a backboard [12-13], or a second reference accelerometer. The latter approach has been applied previously by Aase and Myklebust, as well as by Oh et al. [14-15]. However, in both studies the assumption of full chest decompression during CC was employed.

In this study we present the development of a real-time feedback device for CC during CPR on a soft back support surface, which uses force and dual-accelerometer sensors integrated into a backboard to provide accurate estimation of the CC depth, without assuming full chest decompression.

II. Methods

A. Measurement system and processing

The CPR feedback system being developed at Stellenbosch University consists of two STMicroelectronics MEMS LIS302SG accelerometers (3-axis, ±2 g analog outputs, powered under 3.3 V) which are used to estimate the CC depth during CPR. One accelerometer is mounted on the sternum in a hand pad (connected to the backboard by a bus cable) while the other is embedded in the center of a shutter-ply pine wooden backboard measuring 80 cm x 50 cm x 2.1 cm (length x width x depth), as shown schematically in Fig. 1. To detect complete decompression between each CC it is intended for a compact force sensor to be integrated into the sternal hand pad. For this purpose a flexible force sensor based on a composite piezo-resistive fabric is under development. The system provides active feedback to clinicians during CPR through an array of 15 colored LED lights. The colors: yellow, green and red, indicate respectively a CC depth below, within, or above the recommended CC depth range [16]. An audible metronome is also integrated into the backboard to ensure that the appropriate rate of CC is delivered. The online processing and computation of the CC

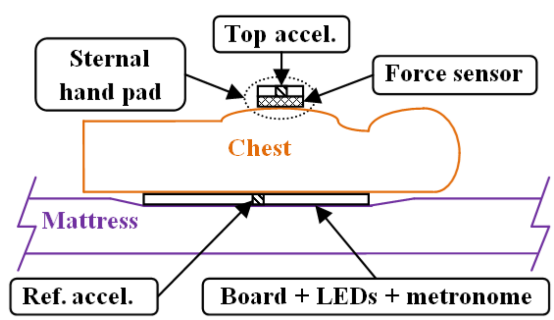

Fig. 1. Schematic of the ‘smart backboard’ CPR feedback system.

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depth using the accelerometer measurements, as well as the control of the LED output display, is performed by an Arduino Nano 3.0 electronic board with an ATmega328 microcontroller.

B. Algorithm for chest compression depth estimation

The CC depth, *d(t)*, is estimated from the dual accelerometer signals by applying two trapezoidal integrations following the algorithm in Fig. 2, based on the equation [15]:

$$d(t) = \iint a(t).dt = \iint (\tilde{a}(t) + \bar{a}).dt = \int v(t).dt$$
$$= \int (\tilde{v}(t) + C_v + \bar{a}.t).dt = \tilde{d}(t) + C_d + C_v.t + \frac{1}{2}\bar{a}.t^2 \quad (1)$$

Where $\bar{a}$ is the acceleration offset (i.e., the remainder after subtraction of the Earth's gravitational acceleration which arises due to imperfect calibration), C_d is the CC depth offset, C_v is the integration constant which is due to arbitrary integration starting time, $\tilde{d}(t)$ is the computed CC depth without an offset, $\tilde{v}(t)$ is the computed velocity without an offset, $v(t)$ is the velocity, $a(t)$ is the acceleration and $\tilde{a}(t)$ is the computed acceleration without an offset.

The first steps (i.e., steps 2 and 14) in the execution of CC depth estimation algorithm involve the projection of the measured acceleration vector onto the continuous component of the acceleration (i.e., the Earth's gravitation). This is done in order to enable the top hand pad to be used in an arbitrary orientation. The key assumption here is that CC is delivered vertically (i.e., parallel to the gravitational direction), which is reasonable in hospital settings, where patients typically lie horizontally on a bed. To discriminate between the continuous acceleration due to the pad orientation and the acceleration due to the periodic CC, a 3rd order Butterworth filter, with a cut-off frequency of 0.1Hz is applied. This choice is based on an empirically determined compromise between the impoverishment of the CC acceleration spectrum and the reaction time needed to detect a change in hand pad orientation.

Execution of the double integration steps in the CC depth estimation algorithm is complicated for two reasons. Firstly, $\bar{a}$ and C_v both make the computed CC depth diverge with time as seen in Eq. (1). Secondly, C_d must be corrected to obtain a physically meaningful measurement (i.e., indicate full chest decompression at a CC depth of 0 mm). The first

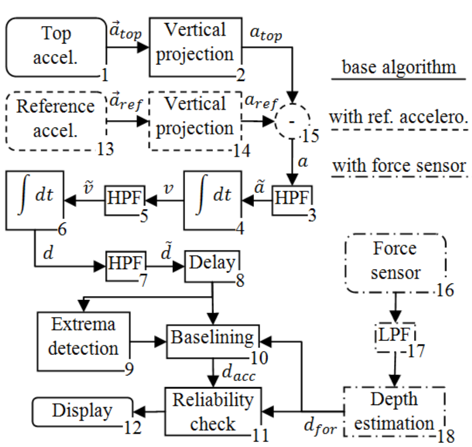

Fig. 2. Algorithm for estimation of the CC depth.

issue is resolved by using 3rd order Butterworth high pass filtering (HPF) to remove offsets and constants, as indicated in steps 3, 5 and 7. The cutting frequencies must be determined with respect to the ERC recommended CC rate range during CPR which corresponds to frequencies between 1.7 Hz and 2.0 Hz [17]. To reduce distortion of the computed CC pattern and to ensure sufficiently fast processing for the feedback application, cut-off frequencies of 0.1Hz are applied in steps 3 and 5, while 1Hz is used in step 7. Although filtering allows for the removal of the continuous components of the signal which cause divergence, it also reduces the amount of information (i.e., harmonics) available in the signal spectrum. The shape of the estimated CC depth as a result does not exactly match that of the actual CC depth.

To compensate for the actual depth offset previous studies baselined *d(t)* by assuming complete decompression between each CC [14-15]. However, it is desirable to avoid making this assumption. One approach to circumvent this is to use a force sensor to detect the degree of decompression (step 18) based on the force-displacement relationship derived from the average chest response determined experimentally by Gruben et al. [17]. Real-time baselining can then be performed by detecting the maxima of $\tilde{d}(t)$ using double derivation and thresholding (steps 9 and 10). To do this the algorithm proposed by Oh. et al. [15] can be applied by detecting the local maxima and setting their actual value. Rather than using preceding and following maxima to set the values in between, only the preceding maximum is used, to avoid delaying the feedback display by a whole CC period. This allows the maxima to be set to correspond with the actual decompression depth, as estimated from the force measurements (d_{for}) rather than arbitrarily setting them to zero (i.e., assuming complete decompression). Also, to ensure that the decompression information is incorporated into the baselining process when computing the CC depth it is necessary to delay the signal by 100 ms in step 8.

C. Experimental Setup

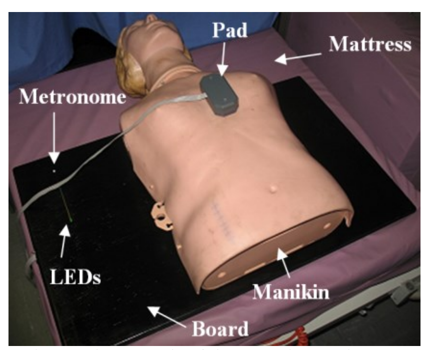

Fig. 3. Photograph of the experimental setup.

The CPR feedback system was tested using the experimental setup shown in Fig. 3. The apparatus consists of an ArjoHuntleigh Contoura 300 series hospital bed, with a Pentaflex mattress from Huntleight Healthcare. All tests were performed using a 3.9 kg Laerdal Little Anne™ Model 020020 torso CPR training manikin measuring 64 cm x 21 cm x 34 cm (height x width x depth). To replicate the precompression of the mattress by the weight of a patient, a 20 kg mass was added to the manikin torso. For the hard back support surface tests, measurements were taken on the floor. The displacement of the manikin chest during CC was measured using a UniMeasure PA-4-CES-R potentiometer, with data recorded at a sampling rate of 1 kHz.

D. *Testing Procedure*

Five tests were conducted to evaluate the performance of the CPR feedback system under various CC conditions (Table I). These included tests on the mattress with no backboard (i.e., hand pad only, Case A), with a backboard without the reference accelerometer (Case C) and with a backboard including the reference accelerometer (Case D). In addition, tests were performed on the floor without the reference accelerometer or backboard using full (Case B) and partial chest decompression (i.e., intentionally incorrect CC, Case E). During each test the following steps were performed:

- Manual CC was delivered by a layman for 60 to 90 seconds at the ERC recommended rate (100 cpm) [16].
- CC depths between 38 mm and 50 mm were applied. (This was done because the manikin was limited to a maximum sternum-to-spine displacement of 52 mm).
- The sternal displacement and hand pad acceleration were measured. (In Case D the backboard acceleration was also measured).

E. *Data Analysis*

The data were imported into MATLAB® (Natick, MA) for offline processing. The actual CC depth was found from the sternal displacement measurements. The CC depth was also estimated off-line using the accelerometer signals based on the algorithm outlined in section B. To make sure the MATLAB® calculations were relevant for real-time computation, the data were processed respecting causality. To match the feedback system's lower sampling frequency of 100 Hz, only one in ten data points (recorded at 1kHz) was used. The data were also re-processed to reflect the lower resolution of the hardware (i.e., 10^{10} bytes rather than 10^{12}). Decompression was detected using the potentiometer measurements and not the force sensor, which is still being developed. Statistical calculations were performed in MATLAB®. Data are presented as mean ± S.D in Table I. Student's t-testing was used to verify statistical significance, with p-values < 0.05 considered significant.

TABLE I
CHEST COMPRESSION RESULTS†

Case	Back support surface	Chest decomp.	Actual CC depth (mm)	Estimated CC depth (mm)	% Difference
A	Mat (no ref)	Full	46.1± 2.6	83.2± 7.0	80.6
B	Floor (no ref)	Full	45.4± 3.4	47.8± 4.4	5.2
C	Mat + BB (no ref)	Full	47.5± 2.3	59.2± 3.9	24.4
D	Mat + BB (ref)	Full	47.5± 2.3	51.0± 7.3	7.3
E	Floor (no ref)	Partial	41.1± 6.7	43.0± 8.3	4.5

BB = backboard; Mat = mattress; ref = reference accelerometer; decomp. = decompression; †T-tests comparing the 5 cases yielded p values < 0.005.

III. RESULTS

Figure 4 shows the differences between the computed and measured CC depths for case D. The solid and dashed lines correspond to the estimated CC depth before and after baselining, respectively, while the dotted line indicates the actual CC depth. The actual and estimated (baselined) mean CC depths for this case are 47.5 ± 2.3 mm and 51.0 ± 7.3 mm respectively, with an average difference of 7.3%.

Figure 5 compares the actual and estimated mean CC depths for the 5 test cases explored. The filled and unfilled bars correspond to actual and estimated CC depth, respectively. The smallest difference between the actual and estimated mean CC depth of 1.9 mm (4.5 %) occurred in case E (floor only, partial decompression), while the largest difference of 37.2 mm (83.2%) was in case A (i.e., mattress only).

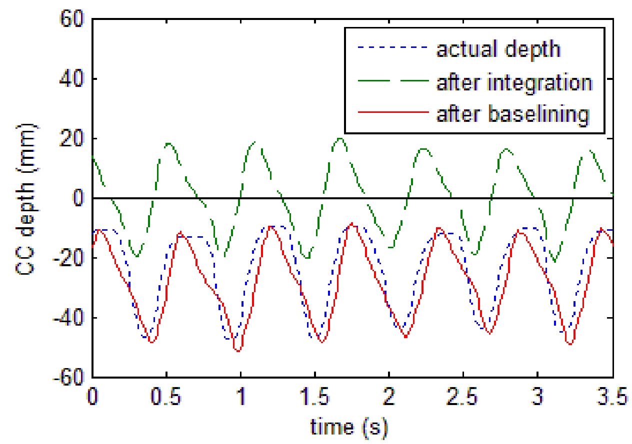

Fig. 4. Accelerometer waveform and its processed signal (Case D).

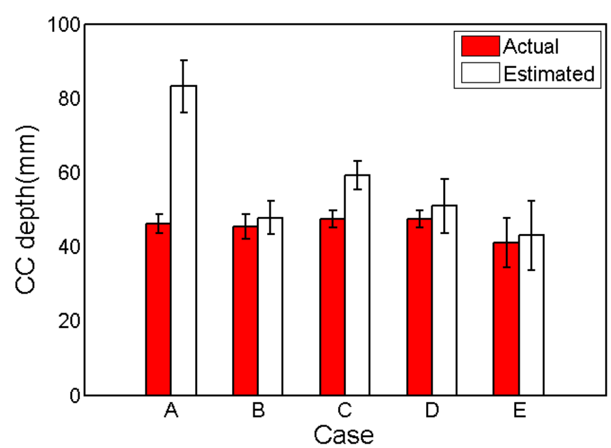

Fig. 5. Actual and estimated CC depths for the five test cases.

IV. DISCUSSION

The results presented in Fig. 4 indicate that the algorithm used by the dual accelerometer CPR feedback system is able to accurately estimate the CC depth once the accelerometer signals are baselined. It is vital to note here that there is a slight time delay (< 0.05s) in the waveform of the estimated CC depth with respect to the actual CC depth due to the speed of accelerometer signal processing. However, this is not expected to be a problem during clinical use of the feedback system since the time lag is very small and will likely be imperceptible to users.

Fig. 5 reveals several useful insights into the performance of the dual accelerometer CPR feedback system relative to that of a single accelerometer system. Comparison of Case A with Case D shows that using a reference accelerometer produces a significantly more accurate estimate of the CC depth. This is highlighted by the fact that the difference between the actual and estimated CC depth in Case A was 83.2% compared to 7.3% in Case D. It is also interesting to compare Cases C and D, which differ only due to the fact that a reference accelerometer was not used in Case C. Although a backboard was used in both cases, there is a significant difference in the accuracy of the estimated CC depth in Case C (24.4%) relative to Case D (7.3%). This result is consistent with the findings of Oh et al. [15] and underscores the unreliability of a single accelerometer feedback system when it is used on a soft back support surface, even in the presence of a backboard. In contrast, on a very hard back support surface, Cases B and E show that a single accelerometer system provides comparable accuracy to a dual accelerometer system (Case D). The impact of partial chest decompression on CC depth estimation can be seen by comparing Case B (full decompression) to Case E (partial decompression). While comparable accuracy is achieved (5.2% vs. 4.5%), the standard deviation in the CC depth increases by 89.8%, from 4.4 mm to 8.3 mm for full and partial decompression, respectively. This implies that incorrect decompression and poor CC technique, may lead to more variability in CC depth, which is consistent with expectation [1].

In addition, it is important to note that the CC depth estimation algorithm presented here differs from previous work [14-15] in its alteration of spectral information to cope with the unstable integration constants, and in its approach to accounting for chest decompression using a force sensor. This latter approach is advantageous since it measures the degree of decompression rather than assuming full chest decompression. Also, the accuracy of the algorithm presented in this study may be improved by the application of more advanced filtering methods, to more efficiently detect and remove the continuous components of the accelerometer signals. One limitation of this study is that the manikin used did not allow CC depths within the ERC recommended range to be achieved. This is not considered to be significant since the main aim of this study is to evaluate the accuracy of the CC depth estimation algorithm.

V. CONCLUSION

This study presented the development of a real-time CPR feedback which uses force and dual-accelerometer measurements to provide accurate estimation of the CC depth on a soft surface, without assuming full chest decompression. Based on adult CPR manikin tests it was found that the accuracy of the estimated CC depth for a dual accelerometer feedback system is significantly better (7.3% vs. 24.4%) than for a single accelerometer on soft back support surfaces, with and without a backboard. From this it can be concluded that a dual accelerometer CPR feedback system can significantly improve the accuracy of CC depth estimation, especially on soft back support surfaces, regardless of the presence or absence of a backboard, when compared to a single accelerometer system. In addition, the manikin tests have shown that the algorithm used is suitable for a real-time, dual accelerometer CPR feedback application since it produces reasonable accuracy in terms of CC depth estimation, even when used on a soft back support surface.

VI. FUTURE WORK

The force sensor will be integrated into the sternal hand pad to enable the detection of chest decompression.

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