Resuscitation 85 (2014) 820–825

Contents lists available at ScienceDirect

Resuscitation

journal homepage: www.elsevier.com/locate/resuscitation

Simulation and education

Development of a real-time feedback algorithm for chest compression during CPR without assuming full chest decompression☆

F. Gohier, K.H. Dellimore, C. Scheffer

Biomedical Engineering Research Group, Department of Mechanical and Mechatronic Engineering, Stellenbosch University, Private Bag X1, Matieland 7602, South Africa

ARTICLE INFO

Article history:
Received 2 November 2013
Received in revised form 5 February 2014
Accepted 7 March 2014

Keywords:
Clinical engineering
Therapeutic and diagnostic systems
Cardiopulmonary resuscitation
Active feedback system
Chest compression
Incomplete chest decompression

ABSTRACT

Objectives: To evaluate the performance of a real-time feedback algorithm for chest compression (CC) during cardiopulmonary resuscitation (CPR), which provides accurate estimation of the CC depth based on dual accelerometer signal processing, without assuming full CDC. Also, to explore the influence of incomplete chest decompression (CDC) on the CC depth estimation performance.

Methods: The performance of a real-time feedback algorithm for CC during CPR was evaluated by comparison with an offline algorithm using adult CPR manikin CC data obtained under various conditions.

Results: The real-time algorithm, using non-causal baselining, delivered comparable CC depth estimation accuracy to the offline algorithm on both soft and hard back support surfaces. In addition, for both algorithms incomplete CDC led to underestimation of the CC depth.

Conclusions: CPR feedback systems which utilize an assumption of full CDC may be unreliable especially in long duration CPR events where rescuer fatigue can strongly influence CC quality. In addition, these systems may increase the risk of thoracic and abdominal injury during CPR since rescuers may apply excessive compression forces due to underestimation of the CC depth when incomplete CDC occurs. Hence, there is a strong need for CPR feedback systems to accurately measure CDC in order to improve their clinical effectiveness.

1. Introduction

Previous work has shown that, even when performed by well-trained clinicians in a hospital setting, the quality of cardiopulmonary resuscitation (CPR) is seldom consistent and often does not meet recommended guidelines.[1,2] This sub-optimal performance is partly due to the poor quality of chest compression (CC) during CPR, i.e., insufficient depth, rate and duration.[3,4]

One strategy for improving the quality of CC during CPR involves the use of a feedback device to assist rescuers in delivering CC at the rate and depth recommended by the European Resuscitation Council (ERC) guidelines.[5] However, there are several challenges which must be overcome in order for a clinically effective feedback device to be developed.

Among the many obstacles involved in realizing an effective feedback device is the need to prevent information overload to the rescuer, since this has been shown previously to lead to poor CC performance during CPR.[6,7] For instance, Oh et al. and Boyle et al.[8,9] have reported that overemphasis on the CC rate may cause clinicians to deliver inappropriate CC depths. However, this latter problem can be easily overcome by using a metronome to guide the frequency of CC.[10] Another major challenge in the development of an effective feedback system relates to the methods used to measure the CC depth and the algorithm applied to process these measurements to provide feedback. This has been previously accomplished by measuring the sternal CC force using devices such as the CPR-Ezy (Health Affairs Ltd., Hertfordshire, UK). However, accurate estimation of the CC depth from force measurements is highly dependent on the patient's thoracic stiffness and damping,[4,11,12] which may limit the feasibility of this approach. Some commercial CPR feedback devices, such as PocketCPR (Zoll Medical Corporation, Chelmsford, MA), avoid this problem by relying on an accelerometer to determine the sternal displacement during CC. This approach works well when CPR is performed on hard back support surfaces, e.g., the floor. However, on soft back support surfaces, such as a bed, the CC depth can be overestimated, thereby making such devices unreliable in these situations.[13] One

☆ A Spanish translated version of the summary of this article appears as Appendix in the final online version at http://dx.doi.org/10.1016/j.resuscitation.2014.03.003.

E-mail address: kiran.dellimore@gmail.com (K.H. Dellimore).

http://dx.doi.org/10.1016/j.resuscitation.2014.03.003
0300-9572/

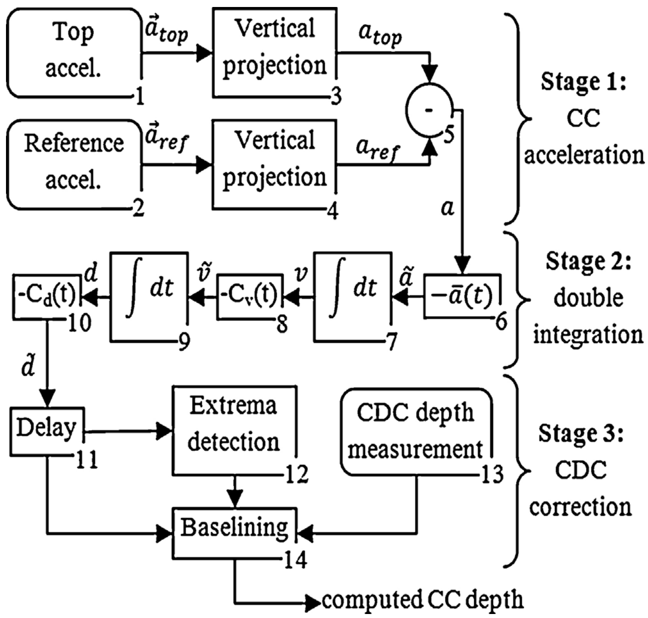

Fig. 1. Method for CC depth estimation.

way to remedy this, as well as to improve the depth of CC is to insert a backboard under the patient during CPR,[14–16] with a second reference accelerometer integrated onto its surface. The use of dual accelerometers has been explored previously in work by Aase and Myklebust,[17] Oh et al.[18] and Gohier et al.[19] However, these studies all utilized the assumption of full chest decompression (CDC), which may lead to suboptimal CC depth estimation, especially in long duration CPR events where rescuer fatigue can strongly influence CC quality.[20,21]

In this study we present a real-time feedback algorithm for CC depth estimation during CPR and evaluate its performance by comparison with an offline algorithm, using experimental CC data obtained under various conditions. The influence of incomplete CDC on the real-time algorithm's CC depth estimation performance is also explored.

2. Methods

2.1. Overview of CC depth estimation algorithm

The CC depth achieved during CPR can be estimated from dual accelerometer measurements in three main processing stages using the algorithm summarized in Fig. 1. Firstly, the accelerometer signals are processed to extract the CC acceleration, i.e., net acceleration of the chest due to compression (steps 1–5). Secondly, the CC depth is estimated by double integration of the CC acceleration (steps 6–10). Thirdly, the CC depth estimate is corrected to account for chest decompression (steps 11–14).

The mathematical double integration of the accelerometer signals (steps 6–10) is complicated by the accumulation of integration errors which must be removed to avoid instabilities in the CC depth estimation. This can be achieved by either detrending or filtering, depending on the time and data available.

Once the CC depth has been estimated from the double integrated CC acceleration it must be corrected to account for the absolute position of the sternum. This is done by offsetting the CDC peaks to match the actual CDC depth. Previous work has utilized either the assumption of full CDC and offset the decompression peaks to zero,[18] or has relied on a switch to determine a decompression threshold and then offset the CDC peaks to zero.[17] However, the present study utilizes the assumption that a decompression sensor could be designed to permit the actual CDC depth to be measured rather than assumed. This sensor could for instance rely on force measurements since previous work has experimentally established the average force-displacement relationship of the human chest during CPR.[22]

2.2. Off-line algorithm

Previous studies have successfully removed the integration errors by taking advantage of the periodic nature of CC during CPR.[17,18] In particular Oh et al. recommend detecting and removing these errors by determining the divergence trends and then subtracting them (steps 6, 8 and 10).[18] However, to accomplish this the entire measurement data set must be available, and this approach is therefore unsuitable for real-time feedback.

Once the integration errors are removed, the CC depth can be accurately estimated. This estimate must then be offset to deduce the absolute CC depth (steps 11–14). An efficient way to do this in an offline context, as reported by Oh et al.,[18] is to compute the affine expression of a line between two CDC peaks. When assuming full decompression this line is subtracted from the CC cycle data to make it fit a constant zero depth. In this study when the CDC depth is available the baseline is set to match the actual decompressions peaks, rather than assuming full decompression.

2.3. Real-time algorithm

When an estimate of the CC depth is needed in real-time, the algorithm presented in the previous subsection is unsuitable. Detrending would necessitate the processing of many CCs before the trends are stable enough to remove the integration errors. Moreover, non-causal baselining would require a significant delay in the outputting of the estimated CC depth due to the determination of the CDC offset. Therefore, in real-time the integration errors are removed by filtering. In practice 3rd order Butterworth high pass filters, with cut-off frequencies of 0.1 Hz for steps 6 and 8 and 1 Hz for step 10, respectively, were sufficient to achieve reasonable accuracy. These frequencies were chosen empirically as a compromise between 'average removal' and impoverishment of the signal's frequency spectrum.

Baselining the filtered CC depth can be successfully performed by modifying the method applied by Oh et al.[18] Rather than determining the offset for each data point by computing the affine relation between the previous and the following CDC peaks, only the previous peak offset was used. The major drawback of this approach is that it is very sensitive to inconsistent CC delivery (i.e., incomplete CDC).

2.4. Experimental setup and testing

The experimental CC data were obtained using the manual CPR simulator shown in Fig. 2. The experimental setup consists of a 3.9 kg Laerdal Little Anne™ Model 020020 torso CPR training manikin measuring 64 cm × 21 cm × 34 cm (height × width × depth) supported on an Arjo-Huntleigh Contoura 300 series hospital bed, with a Pentaflex mattress from Huntleigh Healthcare. To replicate the pre-compression of the mattress by the weight of a patient, a 20 kg mass was added to the manikin torso. 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.

A prototype feedback system, consisting of a backboard and sternal hand-pad instrumented with two STMicroelectronics MEMS LIS302SG accelerometers (3-axis, ±2 g analog outputs, powered under 3.3 V),[19] was also used with the CPR simulator to gather the CC data. One accelerometer is mounted in the sternal