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Virtual arterial blood pressure feedback improves chest compression quality during simulated resuscitation[☆]

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Abstract

Introduction—Quality chest compressions (CC) are the most important factor in successful cardiopulmonary resuscitation. Adjustment of CC based upon an invasive arterial blood pressure (ABP) display would be theoretically beneficial. Additionally, having one compressor present for longer than a 2-min cycle with an ABP display may allow for a learning process to further maximize CC. Accordingly, we tested the hypothesis that CC can be improved with a real-time display of invasively measured blood pressure and with an unchanged, physically fit compressor.

Methods—A manikin was attached to an ABP display derived from a hemodynamic model responding to parameters of CC rate, depth, and compression-decompression ratio. The area under the blood pressure curve over time (AUC) was used for data analysis. Each participant ($N = 20$) performed 4 CPR sessions: (1) No ABP display, exchange of compressor every 2 min; (2) ABP display, exchange of compressor every 2 min; (3) no ABP display, no exchange of the compressor; (4) ABP display, no exchange of the compressor. Data were analyzed by ANOVA. Significance was set at a p -value < 0.05 .

Results—The average AUC for cycles without ABP display was 5201 mmHg s (95% confidence interval (CI) of 4804–5597 mmHg s), and for cycles with ABP display 6110 mmHg s (95% CI of

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Conflicts of interest

None of the authors have a conflict of interest related to this manuscript.

5715–6507 mmHg s) ($p < 0.0001$). The average AUC increase with ABP display for each participant was $20.2 \pm 17.4\%$ 95 CI ($p < 0.0001$).

Conclusions—Our study confirms the hypothesis that a real-time display of simulated ABP during CPR that responds to participant performance improves achieved and sustained ABP. However, without any real-time visual feedback, even fit compressors demonstrated degradation of CC quality.

Keywords

Cardiopulmonary resuscitation; Chest compressions; Simulation; Arterial blood pressure

1. Introduction

The success of cardio-pulmonary resuscitation (CPR) depends on high quality chest compressions (CC).^{1,2} Return of spontaneous circulation (ROSC) after arrest will require reestablished coronary perfusion, which depends on diastolic blood pressure and aortic wall tension during CPR. Paradis et al. showed that a diastolic blood pressure threshold seems to be correlated with increased chances of ROSC in humans.³ Thus, the arterial blood pressure (ABP) becomes an important parameter for effective CC. It has been shown in animal experiments that it takes at least 5 effective CC to reestablish aortic wall tension and overcome vasoplegia after circulatory arrest.⁴ In order to optimize and sustain ABP over time, continuously adjusted CC would be required for the success of CPR, i.e. ROSC. The chance for the CPR provider, the “compressor,” to practically achieve this should be tremendously increased if the information about the accomplished ABP and the trend of it in relation to adjustments of CC was available in real-time.^{5,6} Accordingly, in those environments such as the perioperative setting, that have the availability of invasive blood pressure equipment and competent staff to perform insertion of an arterial cannula, such feedback could be a viable option to better assess the hemodynamic response to CPR and therapeutic measures. Given these environmental conditions, Moitra et al. recommended the placement of an arterial line during acute hemodynamic instability in order to guide therapies in a more accurate and timely manner.⁷ Furthermore, it has been shown recently that any feedback about the performance of CC to the compressor must be considered highly beneficial for directing adjustments to CC.^{8–10} These previous studies focused on single parameters of CC, such as depth of compression, rate, or recoil. Since the achieved ABP during CPR likely depends on the complex hemodynamic response of all of these factors, and since these parameters are taught as components of quality CPR by the American Heart Association (AHA) and International Liaison Committee on Resuscitation (ILCOR), we created a hemodynamic model that can simulate ABP responding to all of these factors during CC. This model was used to test the hypothesis that a competent and physically fit compressor could use such ABP feedback to continuously adjust performance, establishing and sustaining higher ABP, even over a longer period of time. Based on these considerations, this study aimed to test the effect of simulated, real-time ABP feedback and the effect of exchanged or unchanged compressors on the quality of CC during 10 min of simulated CPR.

2. Methods

This prospective study used a Resusci-Anne manikin[®] (Laerdal Medical Corp., Stavanger, Norway) with modifications as described below, in a high-tech simulation environment (MUSC Simulation Center). Direct transmission of CC activity to a linear potentiometer (slide potentiometer – 10 kOhm, 0.5 W; COM-09119) inside the manikin provided a continuous chest wall position signal. A microcontroller (Arduino Uno[®] Inc., Chiasso, Switzerland) passed a 10-Hz processed signal from the potentiometer to a portable computer, which applied a specifically designed mathematical hemodynamic model (see Appendix A) to calculate and display a simulated real-time ABP waveform. The hemodynamic model responded to variations of depth, rate, and ratio of compression to decompression. It also took into account the cumulative effect of several seconds of effective compressions required to prevent an otherwise rapid decay of the pressure and to therefore achieve a maximal ABP.

After IRB approval (Protocol #00017878 – expedited), written informed consent was obtained from twenty ($N = 20$) medical students endorsing more than 5 h of physical workout per week in order to eliminate effects due to variable levels of fitness. Participants received instruction on the proper performance of CPR on the manikin according to AHA/ILCOR guidelines and were encouraged to provide the best achievable chest compressions, but blinded to the hypothesis of the study. Participant compressors all completed each of the following CC sessions:

- (1) Exchange CC every 2 min, no ABP display.
- (2) Unchanged CC, no ABP display.
- (3) Exchange CC every 2 min, with ABP display.
- (4) Unchanged CC, with ABP display.

For the sessions with ABP display, the achieved curve was shown to the compressor on a monitor. The compressors were told to continuously adjust CC in order to achieve the highest BP results. The Y-axis was labeled 0–120 mmHg in 20 mmHg increments; the X-axis was scrolling time (50 mm/s) to replicate current monitors in clinical use at our institution. No further information about waveform-characteristics and no digital numbers were presented. No specific targets were given to the participants beyond generally trying to optimize their compressions. Each session was comprised of 5 cycles of 2 min of CC interrupted for a 10-s simulated pulse check. The manikin was endotracheally intubated; continuous ventilation was simulated. For analysis, the area under the ABP curve (AUC) was derived and saved in repeated integrations of 5-s intervals, thus creating 24 frames within each 2-min cycle and 120 total frames for each 10-min recording.

Since the observed AUC values in this dataset reflected repeated measurements associated with the same rescuer over time and were thus not independent from one another, a general linear mixed model (GLMM) was used to examine the relationships between the experimental groups (the primary independent variable) and the AUC values.¹¹ See

Appendix B for complete description of GLMM model and details of analysis. Data are shown as Mean $\pm$ SD for demographics and as Mean $\pm$ 95 CI for all other data, unless otherwise noted. Significance was defined as $p < 0.05$.

3. Results

A total of 20 participants (15 male, 5 female) were enrolled and completed the study. The average age was 25.0 ± 2.7 years. Participants' average weight, height, and BMI were 76.9 ± 13.4 kg, 178.6 ± 11.6 cm, and 23.9 ± 2.4 kg/m², respectively.

Unadjusted AUC values averaged across participants are illustrated in Fig. 1. As evident from this graph, having an ABP feedback display present resulted in higher average AUC values for both the unchanged and exchanged compressors. The figure also suggests a slight decline in AUC values within the later cycles (i.e. fatigue) when no ABP display is present as opposed to conditions when ABP was displayed. The average AUC for cycles without ABP display was 5201 mmHg s (95% CI 4804–5597 mmHg s), and for runs with ABP display 6110 mmHg s (95% CI 5715–6507 mmHg s) ($p < 0.0001$). The average AUC increase with ABP display for each participant was $20.2 \pm 17.4\%$ 95 CI ($p < 0.0001$).

The GLMM statistical model yielded the following results. First, having an ABP display present resulted in significantly higher AUC values (36 units/frame higher on average, $p < 0.0001$). As seen in Fig. 2, the observed benefit that the ABP display provides is similar for unchanged and exchanged compressor situations ($p = 0.4$).

Although the overall difference in AUC between unchanged and exchanged compressors did not reach statistical significance ($p = 0.5$), the decline in AUC across cycles was significantly greater among unchanged (mean change from cycle 1 to cycle 5 = -1.2 units) versus exchanged (mean change from cycle 1 to cycle 5 = $+12.7$ units) compressors, showing fatigue over the course of 5 cycles ($p < 0.0001$). Similarly, the declines in AUC across frames were significantly greater without an ABP display (mean change from cycle 1 to cycle 5 = -2.0 units) than with an ABP display (mean change from cycle 1 to cycle 5 = $+13.5$ units), demonstrating that the ABP display may retard the effect of fatigue, as shown in Fig. 3 ($p < 0.0001$).

AUC values were not significantly associated with age ($p = 0.14$), height ($p = 0.98$), weight ($p = 0.63$), or BMI ($p = 0.62$); however, in this group of participants, females achieved significantly lower AUC values than males (51 units difference, $p < 0.0001$).

A post hoc analysis focused on whether the participants demonstrated fatigue. Based on within-cycle slopes, 9 subjects were classified as “fatiguers” (see Appendix B); these subjects exhibited a statistically significant decline in AUC values within 1 or more of their 5 CPR cycles. Within the subgroup of 11 “nonfatiguers” the ABP display together with the unchanged CC modus revealed a slightly increased AUC (0.9 mmHg s; $p = 0.9$) in comparison to the exchange modus. All other comparisons in this respect showed greater tendencies of AUC reduction in the continuous versus exchange modus (non-fatiguers without ABP display: 13.2 mmHg s; $p = 0.07$; fatiguers without ABP display: 4.3 mmHg s, $p = 0.7$; fatiguers with ABP display: 5.8 mmHg s; $p = 0.5$).

4. Discussion

Our results confirm the hypothesis that the display of real-time simulated ABP significantly improves the quality of CC in terms of achieved AUC of ABP for both exchanged and unchanged compressors as compared to no visual feedback. Further, although the unchanged compressors did not significantly improve their quality of CC, the availability of the ABP tracing seems to at least delay the effects of fatigue. These results may serve as an encouraging stimulus for further clinical investigations to show if health care providers in high-tech environments should strive for optimized CPR by going beyond published ACLS guidelines to place arterial lines or consciously ensuring the use of any existing arterial waveform during resuscitation of in-hospital cardiac arrest patients.

It has been discussed extensively that any delay or unnecessary interruption of CC becomes critical for ROSC and patient outcome.¹² The current ACLS guidelines take this into account. Those guidelines however are mainly created for laypersons doing CPR, and they are based on a concept of “one-size-fits-all” and keeping it simple. Although it serves perfectly well for the majority, further optimization may be available within inpatient medical environments. For instance, in the perioperative arena, anesthesiologists are more readily available to allow for quicker and earlier endotracheal intubations that can be followed by continuous CC. Technical equipment and competent staff can allow for insertion of an arterial line for ABP feedback to further optimize the continuous CC.

We hypothesized that a competent and physically fit compressor could use ABP feedback to establish and sustain higher ABP, even over a longer period of time. So that we could gain the additional insight that having an a-line feedback is providing about whether it might be considered safe to allow one compressor to go beyond 2 min, we choose a 2×2 design of the study rather than a 1×2 .

Prause et al. provide a case report indicating displayed ABP was feasible in the out-of-hospital CPR setting and successful in a European setting.⁶ Likewise, there are early reports about CPR inside a US hospital employing arterial lines, as well as recently published data on resuscitations with arterial lines in pediatric critical care settings.^{13,14} In spite of these encouraging reports, it has to be recognized that inserting an arterial line includes inevitable infection risks, and the potential of arterial injuries; particularly since inserting an arterial cannula during ongoing CC might not be an easy task.¹⁵ However, this simulation study was considered a first step to safely gather information helping to determine whether the a-line feedback might be beneficial in principle.

The positive effect of feedback during CPR is well known and has been demonstrated in various studies.^{8–10,16} Most of these studies focus on a single aspect of the performance of CC, such as only compression depth or only compression rate, but the challenge of optimal CC lies in simultaneously combining multiple such parameters to achieve the best ABP. To our knowledge, this is the first publication of a prospective CPR study using a hemodynamic model to simulate the ABP waveform as a realistic, real-time, and physiologically complex response to multiple simultaneous CC parameters.

Even with above average self-reported fitness, our compressors showed different degrees of fatigue. Although the precondition for the study participants was the endorsement of at least 5 h of work-outs per week, chosen as such in order to minimize effects due to variable levels of fitness, there was considerable fatigue over time. Whereas 9 of our fit compressors still demonstrated fatigue, 11 of the 20 participants could be identified as “non-fatiguers”. Attempts to correlate the achieved AUC to height, weight, or BMI failed to unmask a correlation, although there was a trend of higher AUC with greater height and weight. An unexpected observation included a significant increase of AUC among the 15 males compared to the 5 females, even after controlling for the above factors. With respect to the limited setting and sample of this post hoc analysis, any definitive conclusion would be premature, although McDonald et al. found similar results.¹⁷

A post hoc analysis revealed that the group of 11 “non-fatiguers” was able, with ABP feedback, to sustain higher ABP over a long period of time (i.e. 10 min) when left unchanged rather than exchanged. While unlikely to translate to the average health care provider or layperson, this supports the idea that compressor selection may play a significant role in CC quality. It may be that the direct feedback from the a-line acts as motivation to maintain best BP results over time (AUC).

Additionally, among our group of generally young and active medical students, the benefit conferred through rotating compressors was less important ($p = 0.5$) than that achieved with the presence of simulated real-time ABP feedback ($p < 0.0001$). On average, our relatively fit group of compressors began to fatigue beyond the second or third cycle of compressions, especially when no ABP feedback was provided (Fig. 3). Fatigue would be expected to be even more pronounced amongst the average, relatively less fit, resuscitation team responder, confirming the ILCOR/AHA guideline for 2-min exchange of the compressor. However, with the importance of CC quality on outcomes¹¹ and such variation in the ability of responders to sustain optimal CCs, some additional consideration should be given to carefully selecting more physically fit persons to optimize CC. It may turn out to be more beneficial for a fit compressor, utilizing a-line feedback, to provide the first 2–3 sequential cycles rather than be replaced by a less fit compressor unable to match the same CC quality.

5. Limitations of the study

The idea of this simulation study was to be a starting point for further investigations. This was a low risk way to find out what would happen if an above average fit compressor was allowed to adjust their CC-performance according to real-time ABP feedback. Rather than an attempt to directly challenge BLS or ACLS guidelines, it is hoped that this study will trigger further clinical investigation with particular focus on in-hospital high-tech environments.

In spite of possible benefits, the placement of an arterial line during CPR would require specially trained performers with excellent skill. Nevertheless, it would include considerable risk of injuries and infection.

The displayed ABP in this study, was dependent on preset hemodynamic model assumptions, and may not accurately represent the “real” blood pressure response during

CPR. However, the model's optimal values were predetermined within a range from 0 to 120 mmHg in order to allow deviations but also to require the compressor to maintain aggressive efforts.

Although hemodynamic modeling is well known,^{18,19} its direct application to simulation settings and equipment is not currently practiced for augmenting CPR training. Previous hemodynamic algorithms may have been impractical for general real-time simulation. Our model provides the real-time ABP feedback needed for immediate fine-tuning of CC, which would likely be helpful in patients with variable pathophysiological conditions. Our hemodynamic model represents only one particular method of translating, via raw data produced from a simple potentiometer, CC into a realistic appearing hemodynamic response. The main advantage of our hemodynamic model lies in its versatility and the possibility to facilitate simulated real-time ABP waveforms with minor hardware modifications to a standard manikin. Modifiable assumptions within our hemodynamic modeling will produce significantly different simulated blood pressure results. Repeating studies including variable modeling to simulate unique patho-physiological preconditions of circulatory arrest could show if our hypothesis holds true under varying situations. Our current fixed arbitrary assumption model primarily strived to account for the dependence of ABP on the complex interaction of measured variation between rate, depth, compression–decompression ratio, and delays.

6. Conclusion

There appears to be an opportunity for improved ACLS within high-tech medical environments (ICU, OR, PACU) by achieving a higher quality of CC using real-time, continuous ABP feedback to the compressor. Insertion of an arterial line during CPR may even be beneficial if competent personnel can accomplish it without adversely impacting the ACLS. Although fatigue happens even to physically fit personnel, it is suggested that compressor choice may be quite important. Since our data suggest that availability of ABP feedback was more important than rotation of compressor to avoid fatigue; with AFP feedback, some particularly strong and fit compressors may safely perform better CC for a longer duration (for up to 10 min in our study) compared to a lesser competent compressor rotating into the role every 2 min. Methods of identifying or training such optimal compressors to provide improved CPR quality as part of inpatient ACLS teams is a promising area for ongoing inquiry.

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Appendix A.

The sensor provides a calibrated number between zero and one corresponding to the depth of chest compression which will be called the forcing f .

To simulate the dynamics of blood pressure in a patient undergoing chest compression, a dampened harmonic oscillator coupled to a decaying reservoir was used. The oscillator's amplitude $\times$ due to the forcing f is described by:

$$\frac{d^2}{dt^2}x + \frac{dx}{dt}y + \left(\omega_r^2 + \frac{\gamma^2}{2}\right)x = f$$

Here, ω_r is the resonance frequency and γ is the damping of the system. Assuming f to be sinusoidal, the oscillator will respond with a phase-shifted periodic signal of the same frequency ω_f .

For times $t > 1/\gamma$, the maximum amplitude will be

$$A(\omega_f) = \frac{F}{\sqrt{\left(\omega_f^2 - \omega_r^2 + \frac{\gamma^2}{2}\right)^2 + \gamma^2\omega_f^2}}$$

x is rescaled to x' such that the amplitude during resonance is equal to one. By choosing γ , the width of the frequency range in which resonance occurs can be set. Since the sensor can only provide positive numbers, the forcing will more likely be of the form $f(t) = 0.5(\sin(\omega_f t) + 1)$. This has been incorporated into the rescaling. Now a second differential equation representing a decaying reservoir is used:

$$\frac{dp}{dt} + \lambda = g(x')$$

Here, p is the resulting blood pressure, $[H9261]$ is the pressure decay frequency and $g(x')$ is defined as:

$$g(x') = \begin{cases} x', & \text{for } x' \geq 0 \\ -\theta x', & \text{else} \end{cases}$$

This is effectively the absolute value of x' with the negative parts multiplied by the filling factor θ . Again, the output p is rescaled such that during resonance the mean pressure is equal to the target pressure, which can be set in the software. λ is used to set the speed at which the blood pressure decays. More precisely, when the chest compression has stopped, p will begin to decay exponentially so that after time $t = 1/\lambda$ the value of p falls off to $p/e \approx 0.368p$ (Fig. A1).

Appendix B.

In the GLMM, AUC is expressed as a function of a number of factors, including compression type (exchanged vs. unchanged), ABP display vs. no ABP display, cycle number (1–5, to account for any fatigue between cycles), frame number (1–24, to account for fatigue within cycles), binary indicator variables for frame number 1 and for frame

number 24 (to account for the pulse checks between cycles), and participant age, gender, and height. In addition to these main effects, several 2- and 3-way interaction terms were also examined to account for as much of the variation in AUC values as possible without significant model over-fit. The final GLMM included the following interaction terms: Compression type $\times$ ABP display presence, compression type $\times$ cycle number, compression type $\times$ frame number, ABP display presence $\times$ cycle number, ABP display presence $\times$ frame number, and compression type $\times$ cycle number $\times$ frame number. The inclusion of interaction terms allow to account for a differential degree of decline within cycles in AUC over time in the “unchanged” setting compared with the “exchanged” setting. In order to account for the within-participant correlation, the GLMM includes random intercepts and slopes for participants. A combination of a compound symmetry and auto-regressive covariance structure was selected based on Akaike Information Criterion [Akaike 1974]. This statistical model tested each of the study hypotheses.

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Fig. 1.

Averaged AUCs of ABP for a CPR session of 10 min. Display of 5 cycles of chest compressions interrupted for 10 s pulse checks. With BP display continuous compressors do slightly better in the first 2 cycles, but become fatigued later; for condition without BP display, exchanged compressors do slightly better (not statistically significant) [AUC = area under the curve; CPR = cardiopulmonary resuscitation; CC = chest compressions; ABP = arterial blood pressure].

Mean AUCs Achieved for Each Frame of Chest Compressions for Each CPR Condition

Fig. 2.

The histogram displays AUCs for the condition of continuous compressor (left) and of exchanged compressors (right) for ABP display (A-Line Present) versus no ABP display (No A-Line Present). ABP display based on a simulated presence of an arterial line contributes to significantly higher AUC under each condition. $N = 20$ for each condition. Data have Mean $\pm$ 95 CI [AUC = area under the curve; CPR = cardiopulmonary resuscitation; ABP = arterial blood pressure].

Fig. 3.

This histogram illustrates fatigue as percentage of change in AUC from baseline for the various CPR conditions. Lack of ABP display together with unchanged CC – even for above average fit personnel – was associated with the worst signs of fatigue. Exchange of compressors together with ABP display was associated with the lowest level of fatigue. The initial frame of the cycle was used as the baseline reference for comparison [AUC = area under the curve; CPR = cardiopulmonary resuscitation; CC = chest compressions; ABP = arterial blood pressure].

Fig. A1.

The figure shows the effect of different values for [H9253] on the amplitude response of the oscillator. Here $\omega_r = 100/\text{min}$.