The cardiac output from blood pressure algorithms trial

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<th>Citation</th>
<th>Sun, James X. et al. &quot;The cardiac output from blood pressure algorithms trial.&quot; Critical Care Medicine. 37(1):72-80, January 2009.</th>
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<tr>
<td>As Published</td>
<td><a href="http://dx.doi.org/10.1097/CCM.0b013e3181930174">http://dx.doi.org/10.1097/CCM.0b013e3181930174</a></td>
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<tr>
<td>Publisher</td>
<td>Wolters Kluwer</td>
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<tr>
<td>Version</td>
<td>Original manuscript</td>
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<td>Accessed</td>
<td>Wed Feb 06 05:44:51 EST 2019</td>
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<td>Citable Link</td>
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# THE CARDIAC OUTPUT FROM BLOOD PRESSURE ALGORITHMS TRIAL

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<td>draft</td>
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<td>Manuscript Type:</td>
<td>Original Articles - Clinical Investigations</td>
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<td>Date Submitted by the Author:</td>
<td>n/a</td>
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<td>Complete List of Authors:</td>
<td>Sun, James; Massachusetts Institute of Technology, Electrical Engineering and Computer Science Reisner, Andrew; Massachusetts General Hospital, Emergency Medicine; Massachusetts Institute of Technology, Health, Science and Technology Saeed, Mohammed; Massachusetts Institute of Technology, Health, Science, and Technology Heldt, Thomas; Massachusetts Institute of Technology, Health, Science, and Technology Mark, Roger; Massachusetts Institute of Technology, Health, Science, and Technology</td>
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<td>Key Words:</td>
<td>cardiac output, non-invasive monitoring, pulse contour, database, arterial blood pressure, thermodilution</td>
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THE CARDIAC OUTPUT FROM BLOOD PRESSURE ALGORITHMS TRIAL

July 2007

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This research was supported by grant R01 EB001659 from the National Institute of Biomedical Imaging and Bioengineering and by the Center for the Integration of Medicine and Innovative Technology.

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Introduction

Cardiac output (CO) is a cardinal parameter of cardiovascular state, and a fundamental determinant of global oxygen delivery. Historically, clinical measurement of CO has been limited to critically-ill patients, using invasive indicator-dilution methods such as thermodilution (COTD). Alternative CO measurement strategies have not been widely accepted in critical care, and outside the intensive care unit, rather imprecise metrics are frequently used to assess CO and circulatory adequacy (e.g. blood pressure, urine output, mental status, etc.).

Throughout the past century (and longer (1)), the premise that relative changes in CO could be estimated by analysis of the arterial blood pressure (ABP) waveform has captured the attention of many investigators. Today, peripheral ABP is routinely available in ICU patients, and non-invasive devices exist to measure peripheral ABP in non-critically-ill populations (2, 3). Tracking changes in CO continuously and non-invasively via ABP waveform analysis may be valuable both within and beyond the ICU setting: such a “vital sign” might be a sensitive and specific indicator of circulatory pathology and useful in optimizing therapies such as volume resuscitation and catecholamine infusions. Yet CO-from-ABP monitoring has not been widely adopted in any clinical setting. One can speculate that this is in part because of inadequate validation.

In this investigation, we established an algorithm-testing dataset -- a subset of the Multiparameter Intelligent Monitoring in Intensive Care II (MIMIC II) database (4) --
containing radial artery waveform data and contemporaneous reference \( \text{CO}_{TD} \) measurements, collected in an ICU patient population during routine clinical operations. We used this dataset to evaluate eight previously reported CO-from-ABP algorithms. Moreover, we are making this dataset publicly available, establishing a standard for the meaningful comparison of different CO-from-ABP algorithms using “real-world” (e.g. not artificially pristine) ICU physiologic data. Vendors and future developers can apply their CO-from-ABP algorithms to this public-access dataset and report how they perform relative to alternative algorithms. The MIMIC II \( \text{CO}_{TD} \)/ABP dataset is an analog of the public access arrhythmia databases that have played an indispensable role in the development, refinement, and – ultimately – widespread acceptance of automated algorithms for electrocardiogram analysis (5).
Materials and Methods

Database development: Our CO\textsubscript{TD}/ABP dataset was extracted from the MIMIC II database (4). The MIMIC II database includes physiologic and wide-ranging clinical data from over 2,500 ICU patients (MICU, CCU, and SICU) hospitalized at the Beth Israel Deaconess Medical Center, Boston, USA between 2001 and 2005. Radial ABP waveform data from the M1006B invasive pressure module and CO\textsubscript{TD} data (temporally resolved to the nearest minute) were originally sourced from Philips CMS bedside patient monitors (Philips Medical Systems, Andover, MA). Waveforms were sampled at 125 Hz with 8 bit resolution. The patients’ gender and age were input by the nursing staff as part of routine clinical operations, using the Philips CareVue system, and these archived data were another component of the MIMIC II database. Additional details about the MIMIC II database are available in (4). A software routine developed using Matlab (Mathworks, Natick, MA) identified and extracted MIMIC II cases with CO\textsubscript{TD} measurements and contemporaneous one-minute-long segments of radial ABP waveform. All Matlab algorithms used in our analyses, and the MIMIC II CO\textsubscript{TD}/ABP dataset, have been contributed to PhysioToolkit and PhysioBank, respectively, and are available for review and free public use from the PhysioNet website (6).

ABP Signal Processing: Within each minute-long ABP segment, individual heart beats were identified using a Matlab implementation of an algorithm by Zong (7). The waveform quality of each ABP pulse was assessed automatically using a signal
abnormality index (SAI) algorithm (8). A set of features from each ABP pulse was computed, including the peak (systolic blood pressure, SBP), trough (diastolic blood pressure, DBP), MAP, and pulse pressure (SBP minus DBP); see Fig. 1. Each ABP pulse's average of negative slopes was computed, a metric of spiky, non-physiologic noise in the ABP pulse waveform. After computing the preceding features for an ABP pulse, the SAI algorithm checks that all were within normal limits (8). The SAI also checks that the features’ variation from one ABP pulse to the next is within normal limits. The SAI algorithm reports a binary ‘normal’ or ‘abnormal’ rating for each ABP pulse, depending if all the normality criteria were met (8). Any abnormal beat was excluded from further analysis. If a given minute-long ABP segment contained more than 40% of abnormal beats, the entire segment (and its corresponding CO_TTD) was excluded from further analysis.

In addition, we estimated the duration of each entire beat and its systolic interval. There is no single widely accepted method to identify the systolic interval in a peripheral ABP pulse (in contrast to a central ABP pulse, the dicrotic notch in a peripheral ABP pulse does not indicate closure of the aortic valve). Therefore, we chose two alternative criteria to identify the end of systole. First, we computed a heuristic estimate of systolic duration, \(0.3 \cdot \sqrt{\text{beat} \_ \text{period}}\), originally suggested as an approximation of the QT interval (9). Second, we identified the point after SBP with the lowest non-negative slope, as in Fig. 1. In practice, this method located the trough of the dicrotic notch, or any relative plateau which persisted for two or more ABP samples.
Investigational CO-from-ABP algorithms:

Mean arterial pressure (MAP) is positively but imperfectly correlated with CO. Of course, variable degrees of systemic vasoconstriction or dilation (which affect peripheral vascular resistance, PVR), as well as variable venous pressure, make MAP an unreliable predictor of CO. ABP waveform analysis assumes that other features in the waveform are less affected by confounders such as peripheral vascular resistance and are thus more reliable correlates of CO. MAP serves as our control method against which eight investigational CO-from-ABP methods are compared. The investigational algorithms are summarized in Table 1. Most algorithms predict stroke volume, and CO is taken as the product of stroke volume and median heart rate over the one minute window. Many of the algorithms were initially intended for a central aortic ABP waveform; in this investigation, we explore their application to a peripheral radial ABP.

(A) Pulse Pressure: In 1904, Erlanger and Hooker suggested that the pulse pressure is a surrogate of stroke volume (10). This notion naturally arises from a basic Windkessel model of the arterial tree, in which the arterial system is considered a single elastic tank, with flow exiting through a distal resistive element. Assuming that forward flow is correlated with pulsatile volume and cardiac ejection is near-instantaneous, then the product of pulse pressure and heart rate is a predictor of CO.
(B) Liljestrand and Zander: Compliance of an artery varies with blood pressure. The Liljestrand algorithm accounts for the dependence of a subject’s compliance on arterial pressure by scaling its CO estimate to the reciprocal of MAP (11).

(C) Systolic Area: A number of methods treat an artery as a long viscoelastic tube, a “transmission line” model. Within a transmission line, pressure gradients accelerate or decelerate flow. By assuming that retrograde (reflected) pressure waves are negligible during systole, it is possible to estimate the pressure gradient and the forward flow from an ABP waveform. Specifically, flow is proportional to the ABP systolic area (the area under the systolic portion of the ABP pulse) (12, 13).

(D) Kouchoukos Correction: A potential source of error is the assumption that cardiac ejection is so rapid that no blood flows out of the arterial tree during systole (“run-off”). Kouchoukos proposed a simple correction factor, related to the ratio of systolic-to-diastolic duration (14); this was a variation of an earlier method proposed by Warner (15).

(E) Diastolic Decay: Bourgeois developed an algorithm to quantify systolic run-off (16). This method leads to an estimation of PVR (CO can then computed from MAP / PVR, assuming CVP is negligible). Bourgeois’ method is based on a constant compliance Windkessel model. In such an idealized model, a mono-exponential diastolic decay (due to the arterial run-off) is expected in the ABP pulse waveform, and that diastolic curve changes solely as a function of PVR (17). Our diastolic decay method
adapts the original Bourgeois method to the radial ABP. We fit a monoexponential curve to just two points of each ABP pulse, taking the peak of systole as the onset of a monoexponential decay, and the trough of diastole as its end (18).

(F) Herd: In theory, systolic blood pressure may be prone to amplification due to early reflected waves. Herd proposed another empirical method, the difference between mean and diastolic pressure, as a more robust surrogate of stroke volume (19).

(G) Corrected Impedance: Wesseling's Corrected Impedance method provides an empiric correction to the systolic area-under-the-ABP curve approach, to account for some of the sources of error described above (20).

(H) AC Power: We explored if the stroke volume bore any reliable relationship with the ABP waveform root-mean-square.

Modelflow, PiCCO, PulseCO, and FlowTrac: The initial investigational plan was to re-implement the best known commercial algorithms, based on publicly-available information (21-27). However, we found that insufficient public information was available to effectively replicate these algorithms. This matter is addressed in detail in the Discussion.
**Calibration:** We applied the investigational algorithms described above and summarized in Table 1 to the data of subjects who have at least two paired measurements of $COTD$ and a contemporaneous, minute-long segment of ABP waveform of sufficient quality. Each algorithm was calibrated to each patient, using two different methods. First, the “best-possible calibration factor” was computed, $C_1$ (fig. 2). $C_1$ was selected to minimize the root-mean-square of the difference of each pairing of $COTD$ and $CO$-from-ABP. Next, each algorithm was calibrated to each patient using a different methodology, $C_2$ (fig. 2). $C_2$ was established only by the first pairing of the CO estimate and $COTD$, so that $COTD_{\text{initial}} = (C_2 \times \text{CO estimate})$. All investigational algorithms were compared against MAP as calibrated predictors of CO. MAP was calibrated exactly like other algorithms, i.e. $C_1$ and $C_2$.

**Statistical analysis:** Each paired $CO$-from-ABP and $COTD$ had an identifiable "error" (their difference). The distribution of errors for each investigational algorithm was computed, for both the $C_1$ and $C_2$ calibration methods. From these error distributions, 95% limits-of-agreement were computed for each $CO$-from-ABP algorithm, per Bland-Altman methodology (28). We tested if the error distribution for an investigational algorithm was statistically different from the error distribution of "calibrated MAP" (using the $C_1$ data), using the Kolmogorov-Smirnov test (in Matlab).
Results

We excluded 13.7% of the available minute-long ABP data segments (and their paired CO\textsubscript{TD} measurements) which did not pass our data quality criteria – those were segments that contained more than 40% of abnormal ABP pulses. Table 2 shows the characteristics of the 120 subjects analyzed. Typical of an ICU population, we studied older subjects (age 69 years ± 12 s.d.), 67% male. The average length of stay in the ICU was slightly over 2 days with an average of ten CO\textsubscript{TD} measurements per patient. On average, each subjects’ CO\textsubscript{TD} varied by ±46%, PVR varied by ±50%, and MAP varied by ±32%.

In Table 3, we tabulate the 95% limits-of-agreement for all investigational algorithms using both the $C1$ and $C2$ calibration methods. The Liljestrand algorithm performed the best. Figure 3 plots the differences (“errors”) between the Liljestrand method and CO\textsubscript{TD}, and subplots in Figure 3 show error plotted as a function of various physiologic parameters. One notable trend is that the Liljestrand error grows larger as peripheral vascular resistance lessens.

The results in Table 3 and Figure 3 are exclusive of the minute-long ABP segments with >40% abnormal beats. After recomputing the Liljestrand 95% limits-of-agreement for all data-segments (i.e. regardless of data quality) the Liljestrand limits-of-agreement grew to -1.88 / +1.57 L/min. By contrast, applying more stringent ABP quality criteria (analyzing minute-long ABP waveform segments with no more than 5%
of abnormal beats), the limits-of-agreement were reduced to -1.48 / +1.29 L/min, although this more stringent ABP quality criteria excluded 40% of the available ABP data segments.

Some of the algorithms require determining the end-of-systole in a radial ABP pulse. The results in Table 3 are all based on our heuristic method $0.3 \cdot \sqrt{\text{beat \_ period}}$. For select algorithms, subscripts in Table 3 report the results for an alternative method of identifying the end-of-systole (the “lowest non-negative slope” method), which trended towards worse results.
Discussion

*A Public Access Database*: The ability to monitor CO continuously by analyzing the ABP waveform, wrote Wesseling in 1983, could be used to provide “an early warning signal if cardiac output would rise or fall suddenly, to adjust drug rates and infusion rates…to sense bleeding…to get a true mean cardiac output under arrhythmias, etc. (20)” Yet ABP waveform analysis has not been broadly adopted. Perhaps this is in part because these algorithms haven't been convincingly validated (29). Such validation could be enabled by one or more publicly available “standards” databases. In the 1970's, this Laboratory made the BIH-MIT Arrhythmia database publicly available (5). The BIH-MIT Arrhythmia database, together with other public access databases, e.g. the European ST-T database, promoted the development of automated ECG interpretation algorithms. 30 years later, computerized ECG interpretation has evolved so that it is now standard in bedside monitors and even automated defibrillators.

Academic and commercial developers can freely access and download the MIMIC II CO\textsubscript{TD}/ABP dataset (www.physionet.org), apply their algorithms, and report their results. This database contains a large number of radial ABP waveforms and contemporaneous measurements of CO\textsubscript{TD} (over 100 subjects and over 1000 paired data points), archived during routine clinical operations. We observe that the typical record shows distinct intervals of relative stability and other intervals of dynamic physiologic change, as in Fig. 4. The range of physiologic states is summarized in Table 2. The data quality in this dataset -- motion artifacts, incidence of dampened catheters, etc. -- are consistent with routine practice, rather than idealized research conditions. To our
knowledge, there is presently no comparable public database. The performance of a novel algorithm, including break-down conditions or generally unsatisfactory performance, can be identified using this testing database. Moreover, the direct comparisons of different algorithms using a standardized testing database may breed healthy competition, and promote clear, iterative improvements. Finally, credible validation using a standard database may encourage adoption of innovative methods by caregivers, particularly when some methods are proprietary and not fully disclosed to the public. The usefulness of ancillary algorithms for CO estimation (e.g. generalized transfer functions to estimate central aortic pressure, or ABP dampening detectors) can likewise be tested.

Because this CO_{TD}/ABP dataset contains “real-world” ICU data, collected during routine operations, the CO reference was a single CO_{TD} measurement. This reflects clinical practice, even though it is an imperfect CO reference method (30, 31). The difference between an investigational algorithm and the CO reference will be increased by errors in either the CO algorithm or errors in CO_{TD}, so the use of single CO_{TD} measurement as a reference may widen the overall limits-of-agreement (Table 3). However, in a dataset this large, it is unlikely that random errors in CO_{TD} measurements will alter whether an individual algorithm is relatively better at predicting CO_{TD} than calibrated MAP, and how it compares relative to other algorithms. Moreover, if and when future investigators collect additional datasets (perhaps using alternative CO reference methods as in (24), or ABP measured from other anatomic locations such as the femoral artery), these datasets can also be freely posted on PhysioNet for public access,
permitting further standardized comparisons of different CO estimators. We suggest that it is beneficial to make available the largest volume of data for the widest range of populations and physiologic states, rather than idealized, smaller datasets.

**Investigational Algorithms:** We investigated eight CO-from-ABP algorithms, many of which were originally intended for use with a central ABP waveform. The notable finding in this trial is the superiority of the CO-from-ABP algorithm described in 1928 by Liljestrand and Zander, which performed significantly better than MAP at predicting changes in CO, and stood out from the other investigational algorithms. The performance of the Liljestrand method is all the more notable because the ABP data were collected during routine ICU clinical operations, during which some degree of motion artifact, catheter damping, improperly calibrated transducers, etc. are inevitable. We employed lenient ABP quality criteria (analyzing all data with $\leq 40\%$ abnormal ABP pulses), which only excluded 13.7% of the noisiest minute-long ABP waveform segments. To the extent that $\text{CO}_{TD}$ is a useful parameter to monitor, the Liljestrand algorithm may enhance standard vital signs. An example of CO estimated continuously by the Liljestrand algorithm versus episodic $\text{CO}_{TD}$ is given in Fig. 4.

We found that the 95% limits-of-agreement between the Liljestrand CO estimates and $\text{CO}_{TD}$ are a function of ABP waveform quality: as the ABP quality criteria are made increasingly stringent, the limits-of-agreement grow tighter, although more data are excluded. Note that we did not explicitly exclude dampened ABP waveforms, aside from requiring that the pulse pressure was $> 20$ mmHg. Waveform damping (due to air
bubbles, thrombus, partial meatus occlusion, etc.) can subtly reduce the measured pulse pressure and so it is a potentially serious source of error for the Liljestrand algorithm which contains pulse pressure in its numerator. All the same, the Liljestrand algorithm is a better predictor of CO\textsubscript{TD} than MAP alone, and appears superior to the other algorithms studied here. If an automated algorithm were able to detect or exclude slightly dampened ABP waveforms, or if the clinical staff took special care to avoid dampened intra-arterial measurements, it is likely that the Liljestrand method, or any of the investigational methods in Table 1, would prove even more accurate.

The modest performance of the other investigational algorithms requires discussion. Factors which may explain these results include: use of real-world ABP waveform data rather than pristine research data (discussed above); use of radial ABP waveforms rather than central ABP waveforms; one-time calibration rather than repeated recalibration; and inclusion of all subjects regardless of cardiac valve function.

Most of the investigational algorithms were originally intended for use with a central ABP waveform, where the systolic interval of the ABP may have relatively fewer retrograde components (i.e., reflected waves). Yet measurement of a radial ABP is a standard clinical practice, where antegrade and retrograde waves are superimposed. Algorithms which perform suitably using a peripheral ABP may prove more valuable because radial ABP is more often available, and because non-invasive devices exist to measure peripheral ABP. Therefore we feel that investigation of these algorithms applied to a radial ABP is warranted. Indeed, we discovered that the Liljestrand method performs
well when applied to radial ABP. There is precedent for applying an algorithm intended for a central ABP on a peripheral BP (23, 24).

Many pulse contour methods prescribe recalibration after each new reference CO measurement, though we did not use this methodology in our investigation. It is intuitive that very frequent recalibration leads to more accurate CO estimations, but extremely frequent re-measurement of $CO_{TD}$ obviates the need for CO-from-ABP algorithms. Statistics based on a recalibration every time the patient’s state changes don’t reveal much about the times when continuous CO-from-ABP algorithms would be most valuable: when a patient’s state is changing and $CO_{TD}$ is not known. Therefore we compared the accuracy of each algorithm through each subject’s range of ICU physiology, employing just a single calibration. The calibration methods in this study included $C1$, the “best possible calibration” (a retrospective construct, in which one optimal calibration factor that minimizes the overall estimation error is employed); and $C2$, in which only the first pairing of CO-from-ABP and $CO_{TD}$ are employed for calibration, and subsequent pairings are examined for accuracy. Presumably, real-world performances will lie somewhere between the ideal of the $C1$ calibration method and the imprecision of the $C2$ method.

We did not exclude subjects based on heart valve function. Rather, we trusted that the ICU staff would only measure $CO_{TD}$ in appropriate patients (e.g. without significant tricuspid regurgitation), and that the ideal CO-from-ABP algorithm would tolerate some aortic valve dysfunction. When we studied the subset of cases with documented normal tricuspid and aortic valve function, we did not find improvement in
any of the algorithms’ performances. Echocardiograms were available in 56 subjects, and normal cardiac valve function was found in 64% of them (36 subjects). For all eight investigational algorithms, the 95% limits-of-agreement with CO\textsubscript{TD} were no better in this subset of 36 subjects with documented normal valve function.

**Commercially-available CO-from-ABP algorithms:** The best known CO-from-ABP algorithms are the commercially-available methods, such as PiCCO, ModelFlow, and LiDCO (21-27), which are analytically more complicated than our investigational algorithms in Table 1. In fact, our initial investigational plan was to also re-implement these commercial algorithms, based on publicly-available information (21-27). However, it became apparent it was not feasible to re-implement these commercial methods. Without inspecting the source code of any algorithm, there are many steps (pre-processing, error and outlier trapping, temporal averaging, and many other specific computational steps) that cannot be meaningfully replicated using just the published reports. Our initial attempts to re-implement these algorithms relied on our own judgments whenever the published sources had ambiguous methodological detail, but these implementations did not perform well (18). Our good-faith efforts to re-implement these commercial methods are available for review at www.physionet.org; needless to say, they should not be construed as equivalent to what is deployed in the commercial products.

Ideally, vendors would make the source code for their methods available for public inspection, to bring unreliable methods to light and accelerate acceptance of
rigorous algorithms. However, this is simply not standard practice for commercial biomedical algorithms. Because most commercial algorithms will remain proverbial black-boxes to the user community, standard testing databases are all the more essential. Indeed, testing proprietary ECG arrhythmia detection algorithms using standard testing databases is a mandatory step in obtaining US FDA approval (5). The MIMIC II $\text{COTD/ABP}$ dataset is now publicly and freely available. We invite developers and vendors of CO-from-ABP algorithms to test their methods and report their performance on this dataset.

**Conclusion**

Public-access testing databases are necessary to determine which CO-from-ABP algorithms are better than alternatives. Such databases can catalyze the development of, and public acceptance of, increasingly effective algorithms. Regarding the set of algorithms that we applied to this testing database, the Liljestrand method is a better predictor of CO than calibrated MAP in an ICU population, using typical ICU radial ABP, after filtering out 12% of the noisiest data. It is even more accurate given the cleanest ABP waveforms. The Liljestrand predictor may be a valuable parameter for intelligent monitoring algorithms when a patient's radial ABP is measured. Future work should examine the sensitivity and specificity of alarm algorithms which use this predictor, versus those that use conventional clinical parameters, and establish outcomes benefits. The corrected impedance algorithm also surpassed MAP as a predictor of CO changes. Other investigational algorithms failed to surpass calibrated MAP.
Acknowledgements

This research was supported by grant R01 EB001659 from the National Institute of Biomedical Imaging and Bioengineering and by the Center for the Integration of Medicine and Innovative Technology (CIMIT).
References


17. Cardiovascular Mechanics [http://ocw.mit.edu/OcwWeb/Health-Sciences-and-
Technology/HST-542JSpring-2004/Readings/]
18. Sun JS: Cardiac output estimation using arterial blood pressure waveforms.
19. Herd JA, Leclair NR, Simon W: Arterial pressure pulse contours during
cardiac output determination by pulse-contour analysis during hemodynamic
22. Joeken S, Fahle M, Pfeiffier UJ: Devices for in-vivo determination of the
compliance function and the systemic blood flow of a living being. In. USA:
Pulsion Medical Systems AG; 2001.
pressure in humans using a nonlinear, three-element model. J Appl Physiol 1993,
74(5):2566-2573.
24. Jellema WT, Wesseling KH, Groeneveld AB, et al: Continuous cardiac output in
septic shock by simulating a model of the aortic input impedance: a comparison
25. LiDCO plus hemodynamic monitor. In., vol. 2006: LiDCO Cardiac Sensor
Systems.
27. Godje O, Hoke K, Lamm P, Schmitz et al: Continuous, less invasive,
hemodynamic monitoring in intensive care after cardiac surgery. Thorac
29. Heldt T: Continuous blood pressure-derived cardiac output monitoring--should
30. Jansen JR: The thermodilution method for the clinical assessment of cardiac
32. Greenfield JC, Jr., Fry DL: A critique: relationship of the time derivative of
flow in man by the computed pressure derivative method. J Appl Physiol 1971,
31(5):792-795.
Fig. 1: Five examples of ABP waveforms and their key features as identified by our automated algorithms. The horizontal axis is sample number, with 125 samples = one second. **Onset** point of each beat is indicated by a black asterisk “*”; **end of systole**, estimated by $0.3 \cdot \sqrt{\text{beat}_3 \cdot \text{period}}$, is indicated by “X”; **end of systole**, estimated by the ‘lowest non-negative slope’ method, is indicated by a “0”. The algorithms also identify systolic (peak), mean, and diastolic (trough) blood pressures (not illustrated). A signal abnormality index (8) checks that all were within physiologic limits and also checks that the features’ variation from one ABP pulse to the next is not excessive.
Fig. 2: Each investigational algorithm required calibration for each subject. We studied two methods of calibrating the algorithms for a subject. In \( C1 \) (see below, black line), the “best-possible calibration factor” was computed. \( C1 \) was selected to minimize the root-mean-square of the difference of each pairing of \( \text{CO}_{\text{TD}} \) and the corresponding \( \text{CO}\)-from-ABP estimations. In \( C2 \) (see below, grey line), the first pairing of the \( \text{CO}\)-from-ABP estimate and \( \text{CO}_{\text{TD}} \) (see below, stem plot) was used to establish the calibration factor, which was then used for all subsequent CO estimation. Note that \( \text{CO}_{\text{TD}} \) is only measured at discrete intervals, while \( \text{CO}\)-from-ABP is continuous (assuming ABP is measured continuously).
Fig. 3: (a) Bland-Altman plot comparing CO estimated by the algorithm (Liljestrand, using the C1 calibration methodology) with \( \text{CO}_{\text{TD}} \). 95% limits-of-agreement for this algorithm and the other investigational algorithm are summarized in Table 3. (b) Liljestrand algorithm error as a function of several variables. Each variable is then put into 3 bins of equal quantity. Rectangular bars represent 95% limits-of-agreement for each bin. For example, as shown, CO estimation error decreases as PVR increases.
Fig. 4: Example of continuous CO-from-ABP estimated by the Liljestrand algorithm (grey line) versus episodic thermodilution CO measurements (stem plots) for a subject over a 50 hour time interval, using a single calibration ($C_1$, see text for details). Pulse pressure (PP), mean arterial pressure (MAP) and heart rate (HR) through this same temporal window are also shown.
Table 1: Investigational CO-from-ABP algorithms
( CO = Stroke volume * HR )

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<th>Algorithm Description</th>
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<tr>
<td><strong>Pulse Pressure (10)</strong></td>
<td>Stroke volume = k ((SBP - DBP))</td>
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<tr>
<td><strong>Liljestrand (11)</strong></td>
<td>Stroke volume = (k \frac{(SBP - DBP)}{(SBP + DBP)})</td>
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<tr>
<td><strong>Systolic Area (12, 13)</strong></td>
<td>Stroke volume = (k \int_{Systole} ABP(t) dt)</td>
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<td><strong>Systolic Area with Kouchoukos Correction (14)</strong></td>
<td>Stroke volume = (k \left(1 + \frac{\text{Duration}<em>{\text{Systole}}}{\text{Duration}</em>{\text{Diastole}}}\right) \int_{\text{Systole}} ABP(t) dt)</td>
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<tr>
<td><strong>Diastolic Decay [adopted from (16)]</strong></td>
<td>Solves for beat-to-beat PVR, fitting a monoexponential curve to each ABP pulse’s peak of systole and trough of diastole, where: (P_{\text{Diastolic}} = P_{\text{Systole}} \cdot e^{-k\text{time}/PVR})</td>
</tr>
<tr>
<td><strong>Herd (19)</strong></td>
<td>Stroke volume = (k (MAP - DBP))</td>
</tr>
<tr>
<td><strong>Corrected Impedance (20)</strong></td>
<td>Stroke volume = (k (163 + HR - 0.48 \cdot MAP) \int_{\text{Systole}} ABP(t) dt)</td>
</tr>
<tr>
<td><strong>AC Power (Root Mean Square)</strong></td>
<td>Stroke volume = (k \sqrt{(ABP(t) - MAP(t))^2})</td>
</tr>
</tbody>
</table>
Table 2: Subject and data characteristics

<table>
<thead>
<tr>
<th></th>
<th>units</th>
<th>mean ± sd</th>
<th>N</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Study Population</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Age</td>
<td>yr</td>
<td>69±12</td>
<td>120</td>
</tr>
<tr>
<td>Stay duration</td>
<td>day</td>
<td>2.3±2.2</td>
<td>120</td>
</tr>
<tr>
<td>(\text{CO}_T\text{D} ) measurements per patient</td>
<td></td>
<td>10±8</td>
<td>120</td>
</tr>
<tr>
<td>(\text{CO}_T\text{D} ) range per patient</td>
<td>L/min</td>
<td>2.3±1.2</td>
<td>120</td>
</tr>
<tr>
<td>MAP range per patient</td>
<td>mmHg</td>
<td>24±10</td>
<td>120</td>
</tr>
<tr>
<td>PVR range per patient</td>
<td>mmHg·s/ml</td>
<td>0.5±0.3</td>
<td>120</td>
</tr>
<tr>
<td><strong>Pooled Data</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(\text{CO}_T\text{D} )</td>
<td>L/min</td>
<td>5±2</td>
<td>1164</td>
</tr>
<tr>
<td>MAP</td>
<td>mmHg</td>
<td>75±10</td>
<td>1164</td>
</tr>
<tr>
<td>Heart rate</td>
<td>bpm</td>
<td>88±17</td>
<td>1164</td>
</tr>
<tr>
<td>PVR</td>
<td>mmHg·s/ml</td>
<td>1±0.4</td>
<td>1164</td>
</tr>
</tbody>
</table>
Table 3: 95% limits-of-agreement between thermodilution CO and investigational CO-from-ABP algorithms

<table>
<thead>
<tr>
<th>Investigational Predictors of CO\textsubscript{TD}</th>
<th>&quot;Best Possible Single Calibration&quot; (C1)</th>
<th>KS test vs. MAP</th>
<th>&quot;First Pairing Calibration&quot; (C2)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lower/Upper Limits (± L / min)</td>
<td>(p-values)</td>
<td>Upper/Lower Limits (± L / min)</td>
</tr>
<tr>
<td>Liljestrand</td>
<td>-1.76 / +1.41</td>
<td>0.0001*</td>
<td>-2.81 / +2.04</td>
</tr>
<tr>
<td>Corrected Impedance</td>
<td>-1.91 / +1.57</td>
<td>0.009 *</td>
<td>-3.39 / +2.28</td>
</tr>
<tr>
<td>Pulse Pressure</td>
<td>-2.07 / +1.73</td>
<td>&gt; 0.05</td>
<td>-3.05 / +2.76</td>
</tr>
<tr>
<td>Systolic Area</td>
<td>-2.07 / +1.73</td>
<td>&gt; 0.05</td>
<td>-2.85 / +3.05</td>
</tr>
<tr>
<td>Systolic Area with Kouchoukos Correction</td>
<td>-2.08 / +1.71</td>
<td>&gt; 0.05</td>
<td>-3.20 / +2.89</td>
</tr>
<tr>
<td>AC Power (RMS)</td>
<td>-2.09 / +1.73</td>
<td>&gt; 0.05</td>
<td>-3.12 / +2.78</td>
</tr>
<tr>
<td>Diastolic Decay</td>
<td>-2.23 / +1.77</td>
<td>&gt; 0.05</td>
<td>-3.22 / +2.57</td>
</tr>
<tr>
<td>Mean Arterial Pressure</td>
<td>-2.20 / +1.82</td>
<td>&gt; 0.05</td>
<td>-3.19 / +3.42</td>
</tr>
<tr>
<td>Herd</td>
<td>-2.66 / +1.89</td>
<td>&gt; 0.05</td>
<td>-3.65 / +3.16</td>
</tr>
</tbody>
</table>

C1 calibration represents a theoretical, retrospective calibration for each algorithm that best matches a subject’s set of CO estimates and paired set of thermodilution CO data. For each investigational algorithm, the distribution of errors (estimated CO minus thermodilution CO) was compared with the distribution of errors of “calibrated MAP” using the Kolmogorov-Smirnov test. C2 calibration uses the first paired CO estimate and thermodilution CO measurement to calibrate each algorithm to each subject, and applies that calibration to all subsequent pairings.

* p less than or equal to 0.05.

# When using the alternative “lowest non-negative slope” method to estimate the systolic interval, the lower/upper limits are -1.94 / +1.54 L/min. Results in Table 3 employ $0.3 \cdot \sqrt{\text{beat} \_ \text{period}}$ to estimate systolic interval.

## When using the alternative “lowest non-negative slope” method to estimate the systolic interval, the lower/upper limits are -2.40 / +1.97 L/min.