

Hypoglycemia Detection in Critical Care Using Continuous Glucose Monitors: An *in silico* proof of concept analysis

Christopher G. Pretty¹, J Geoffrey CHASE², Aaron LE COMPTE², Geoffrey M. SHAW³ and Matthew SIGNAL¹

¹

Department of Mechanical Engineering, University of Canterbury

²

PhD, Department of Mechanical Engineering, University of Canterbury

³

MbChB, FJFICM, Department of Intensive Care, Christchurch Hospital,
Christchurch School of Medicine and Health Science, University of Otago

Work performed at:

- Department of Mechanical Engineering, University of Canterbury
- Christchurch School of Medicine and Health Sciences, University of Otago

Address for Reprints: Reprints available from
Prof J. Geoffrey Chase,
Department of Mechanical Engineering
University of Canterbury,
Private Bag 4800
Christchurch
New Zealand
Email: geoff.chase@canterbury.ac.nz

Financial Support: New Zealand Tertiary Education Commission.

Keywords: Hypoglycemia, CGM, continuous glucose monitor, alarm, glycemic control, blood glucose, sensor

Abstract

BACKGROUND: Tight glycemic control (TGC) in critical care has shown distinct benefits but also been proven to be difficult to obtain. The risk of severe hypoglycemia (< 40 mg/dL) has been significantly increased in several, but not all, studies, raising significant concerns for safety. Continuous glucose monitors (CGMs) offer frequent measurement and thus the possibility of using them for early detection alarms to prevent hypoglycemia.

METHODS: This study uses retrospective clinical data from the SPRINT TGC study covering 7 patients who experienced severe hypoglycemic events. Clinically validated metabolic system models are used to recreate a continuous blood glucose profile. In silico analysis is enabled by using a conservative single Gaussian noise model based on reported CGM clinical data from a critical care study (MAPE 17.4%). A novel median filter is implemented and further smoothed with a LMS fitted polynomial to reduce sensor noise.

Two alarm approaches are compared. An integral based method is presented that examines the area between a pre-set threshold and the filtered simulated CGM data. An alarm is raised when this value becomes too low. A simple glycemic threshold method is also used for comparison.

To account for random noise skewing the results, each patient record is Monte Carlo simulated 100 times with a different random noise profile for a total of 700 runs. Different alarm thresholds are analysed parametrically. Results are reported in terms of detection time before the clinically measured event, and any false alarms. This retrospective clinical data was used with approval from the NZ South Island Regional Ethics Committee.

RESULTS: The median filter reduces MAPE from 17.4% (SD 13%) to 9.3% (SD 7%) over the cohort. For the integral based alarm, median per-patient detection times ranged, t , from -35 minutes (before event) to -170 minutes, with 0-2 false alarms per patient over the cohort and different alarm parameters. For a simple glycemic threshold alarm (3 consecutive values below threshold) median per-patient alarm times were -10 to -75 minutes, false alarms were 0 to 7, but in one case 5 of 7 subjects never alarmed at all despite the hypoglycemic event.

CONCLUSIONS: A retrospective study used clinical hypoglycemic events from a TGC study to develop and analyse an integral based hypoglycemia alarm for use in critical care TGC studies. The integral based approach was accurate, provided significant lead time before a hypoglycemic event, alarmed at higher glycemic levels, was robust to sensor noise, and had minimal false alarms. The approach is readily generalisable to similar scenarios and the results would justify a pilot clinical trial to verify this study.

1.0 Introduction:

Critically ill patients often experience stress-induced hyperglycemia and high levels of insulin resistance, even with no prior diabetes [1-7]. Hyperglycemia worsens outcomes, increasing the risk of severe infection [8], myocardial infarction [1], and critical illness such as polyneuropathy and multiple organ failure [7]. The occurrence of hyperglycemia, particularly severe hyperglycemia, is associated with increased morbidity and mortality in this group of patients [1, 3].

Some studies have shown that tight glucose control (TGC) reduced intensive care unit patient mortality by 45% following control limits of 110 to 140 mg/dL [7, 9-11]. However, there is a little agreement on what constitutes desirable glycemic performance [12-14], particularly with regard to how TGC affects outcome. Thus, despite the potential, many intensive care units do not use fixed protocols [4, 12, 13, 15, 16].

Overall, any glycemic control protocol must reduce elevated blood glucose levels with minimal hypoglycaemia. Thus, minimising risk in the presence of significant variability in insulin resistance resulting from conflicting drug therapies and dynamically evolving physiological condition among others. As patient condition evolves, particularly acutely, TGC and intensive insulin therapy can prove difficult. Protocols or clinical practises that utilize large insulin doses can thus suffer from high glycemic variability and excessive hypoglycemia [17]. As a result, several clinical trials have not achieved the benefit of TGC [17-20].

Hence, there is a significant difficulty in providing protocols that simultaneously provide good performance and TGC without excessive hypoglycemia. The two major reasons or causes of hypoglycemia are often reported to be clinical error and, or combined with, infrequent measurement using bedside glucometers or blood gas analysers [19, 21-24]. Thus, the use of continuous glucose monitors (CGMs) with their rapid 2-5 minute measurement rates offers the opportunity to better monitor patients so that hypoglycemia could be avoided, mitigating this risk significantly.

Typically, in most ICU studies blood glucose is measured 1-4 hourly, faster only if the levels are already hypoglycemic. The result can be very variable glycemic control, especially with longer measurement intervals [25]. Thus, CGMs would also provide the potential to better or more tightly control glycemic levels minimising variability, which has also been strongly linked with mortality, independent of glycemic levels, in these cohorts [26, 27].

However, there have been relatively few successful investigations of CGMs in critical care use [28], although they are well studied in Type 1 diabetes [29, 30]. In particular, one set of TGC trials using them was not particularly successful [31, 32]. They offer the tradeoff of sometimes significant added sensor noise with their far higher, automated sampling rate [28, 33]. However, these sensors and their algorithms are improving every year so the technology is in a state of constant evolution. Hence, their eventual effective use is potentially inevitable and will free clinicians to provide tighter control in the face of highly dynamic metabolic behaviour.

This paper uses data from the SPRINT TGC study [11]. It examines each of the 7 cases of hypoglycemia that was not a function of sensor error. Conservative sensor noise based on reported data in the literature is added to model results fitted to the data to simulate the use of CGM. A novel integral-based algorithm with a median filter is used to develop a robust and readily generalised alarm approach and prove the concept in this in silico study.

2.0 Subjects and Methods:

2.1 Subjects:

This research was conducted as a retrospective study using records from 7 patients admitted to the Christchurch Hospital ICU between 2005 and 2007. Patients were included if they had one or more severe hypoglycaemic episodes ($BG < 40 \text{ mg/dL}$) while on the SPRINT glycaemic control protocol [11]. Patients were excluded if the hypoglycaemic episode appeared to be due to sensor failure [34] or were found to be due to a recording error. Details of the cohort are shown in Table 1.

Table 1: Cohort details, presented as median [100% range] where applicable. APACHE II – Acute Physiological and Chronic Health Evaluation 2. ROD – Risk Of Death.

N	7
Male/Female	4/3
APACHE II score	25
	[12-30]
APACHE II ROD (%)	53
	[5-72]
Age (yrs)	63
	[37-81]
Hypoglycemic blood glucose level (mg/dL)	38
	[31-40]

The requirement for patients in this study to be on the SPRINT protocol ensures they have regular, consistent and accurate records of blood glucose level (1-2 hourly), and insulin and nutrition administration [11]. The use of these patient records falls under existing ethics approval granted by the Upper South Regional Ethics Committee, New Zealand.

2.2 Methods – CGM Noise Model:

CGM sensor error consists of a bias due to calibration drift with a random, or quasi-random noise superimposed on top. Calibration drift due to sensor degradation over time was not considered in this study as this is controlled by the specific calibration protocol used with the sensor. The random component of noise modeled in this study was assumed to be independent and normally distributed.

This research used a single-Gaussian noise model. The model was derived to produce similar errors on a similar cohort to those reported by Goldberg et al in a 2004 study of the Medtronic CGMS (Minimed-Medtronic, Northridge, CA) in a medical ICU [28]. In the study by Goldberg et al, the calibration of the CGM sensors was performed retrospectively with all the available data and at least 4 reference BG measurements per day [28], removing any bias. This report was used as it was critical care specific and reported a wide range of error statistics versus a reference measure. Goldberg et al reported errors for measurements in 5 BG ranges. The noise model was thus created to match the reported noise statistics over the same 5 ranges.

More specifically, it is a Gaussian distributed random error with a mean absolute percentage error (MAPE) of 12.8% and standard deviation (SD) of 10% on a cohort similar to Goldberg's. However, the percentage error was greater for lower blood

glucose levels here (MAPE 17.8%), given the relatively constant mean absolute difference (MAD) in mg/dL seen. Since, such lower measurements are more frequently encountered in this study involving tightly controlled patients on SPRINT, and which are limited to the time periods around hypoglycaemic episodes, the errors in MAPE are conservatively higher. However, these errors are also more relevant as they are specific to the hypoglycemia alarm situation studied here.

2.3 Methods – In Silico CGM Measurements:

Using a model derived from the clinically validated glucose-insulin model of Lin et al [35], and hospital records, a blood glucose profile was generated for each patient at 5 minute intervals, based on their model-fitted time-varying insulin sensitivity. These profiles were generated in a data window starting 9 hours before the severe hypoglycemic event at a normoglycemic level and ending 4 hours after an actual measured hypoglycaemic event. Random noise was added to this ‘actual’ blood glucose profile using the single-Gaussian model, creating a sequence of virtual CGM sensor outputs.

2.4 Methods –CGM Filtering:

To simulate the real-time use of a CGM device, an algorithm was implemented in Matlab™ (The Mathworks, Natick, MA) to step through the sequence of virtual CGM readings and filter them without knowledge of ‘future’ values, as would be the case clinically. Thus, the clinical situation can be simulated for developing an alarm methodology.

A combination median filter and least mean squares (LMS) curve fit was used to smooth the noisy virtual CGM sequence. Initially, a weighted median filter was applied to the prior 30 minutes noisy data, followed by a linear least squares fit over the previous hour median filtered data.

The fundamental steps to implement the median filter at any given time $t = x$, are:

1. Take current and 2 prior CGM readings (3 sample, 10 minute window) and find median value M_3
2. Take current and 6 prior CGM readings (7 sample, 30 minute window) and find median value M_7
3. Take average of M_3 and $M_7 = M_A$
4. Take current M_A value and 12 prior M_A values (1 hour window of median filtered values)
5. Fit LMS 1st order polynomial line
6. Output value at time $t = x$ is the value of this fitted line at $t = x$

This set of steps is based on well known median filtering [36] and LMS polynomial fitting methods. The multiple windows give an empirically designed trade-off between fast dynamics and response, and longer windows and smoother filtered outputs with lag [37]. This approach is also less computationally expensive than integral based approaches or Kalman filtering [38-40].

2.5 Methods –Alarm Design:

An algorithm was required to trigger an alarm when the filtered blood glucose sequence appeared to be heading towards a hypoglycaemic event. While better than the raw data, the filtered data was still too noisy to apply a simple set of conditions such as m measurements below a threshold BG value. Therefore, a windowed

integral method (essentially an FIR filter [41]) was implemented using the filtered data. This approach is both simple and robust to noise.

Specifically, the area between the BG curve and a specified level was calculated within a window of prior samples. When this integral became less than a pre-selected threshold value, an alarm was triggered, indicating an impending hypoglycaemic episode. Several combinations of these parameter values were simulated. Figure 1 shows an example.

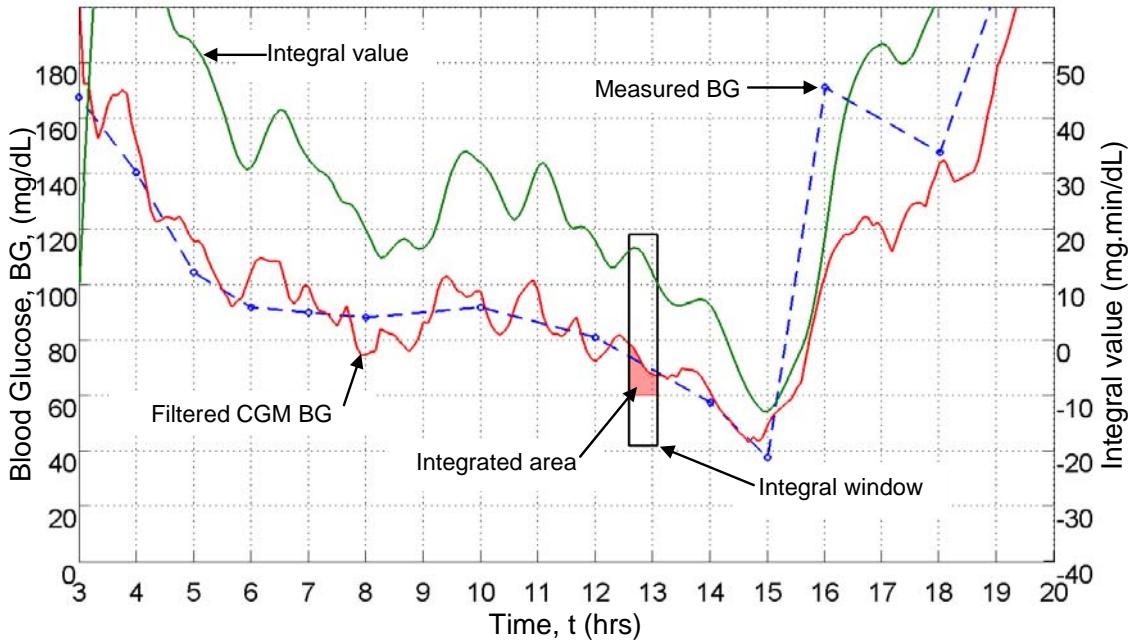


Figure 1: An example showing the blood glucose integral used to trigger a hypoglycaemic alarm.

An alarm was considered false if the following conditions held:

- There was more than 1 alarm for each actual hypoglycaemic episode per patient AND
- For 2 clinical blood glucose measurements either side of the alarm, no value was less than or equal to 40 mg/dL.

It should be noted that this use of prior knowledge of clinical BG measurements was only used after the filtering and processing to identify any false alarms, and would thus not be a part of a real-time implementation.

2.6 Methods –Analysis:

For analysis, the timing of this alarm was compared with the time that the episode was actually detected in the hospital. This value essentially measures the lead time to intervene and the minimisation of minor or moderate hypoglycemia. The number of false alarms was also recorded to ensure that the method was accurate.

To get meaningful results with the random noise, this study used a Monte Carlo analysis approach. Each patient's model-based, true BG profile was passed through the single-Gaussian random noise generator 100 times, creating 100 different virtual CGM sequences per patient. The results of these 700 trials with the filter and alarm

algorithm were analysed using non-parametric statistics to determine overall cohort and per-patient results.

Results were reported in terms of alarm lead time and number of false alarms for several analyses and thresholds. A simple glycemic threshold method is also shown for comparison.

3.0 Results:

3.1 Filter Results:

The median and LMS based filter design resulted in a significant noise reduction. Specifically the MAPE on raw noisy virtual outputs was reduced from 17.4% (SD 13%) to 9.3% (SD 7%) over the cohort. However, given relatively smaller numbers of hours ($N = 91$ hours over 7 patients) the distributions were not perfectly normal. Thus, the non-parametric filtering output was a reduction from a median APE of 14.4% [IQR: 6.8 - 24.9] to a median APE of 7.6% [IQR: 3.6 - 13.2]. Mean absolute differences (MAD) were 14.7 mg/dL and 7.7 mg/dL, respectively. All these values compare well with results reported in the literature [28, 39, 40].

3.2 Alarm Analysis Results:

The primary result for this study was the time difference between an alarm triggered by the simulated CGM data and when a hypoglycaemic episode was detected by actual measurement. As a secondary result, the number of false alarms triggered is also reported as a measure of the alarm algorithm's reliability. Table 2 shows results for the integral-based alarm for different window lengths and trigger threshold values. Negative time values indicate that an alarm was triggered before a hypoglycaemic event was measured. These results are counted over all 700 Monte Carlo simulation runs.

Table 2: Early detection of hypoglycaemic episodes reported for the integral-based alarm algorithm with a range of parameter values. Data are median [IQR] for all 700 Monte Carlo runs (100 per patient).

Integration window length (samples)	5			7			13		
Integral threshold (mg.min/dL)	5	10	20	5	10	20	5	10	20
Hypo detection time cohort (min)	-65 [-110, -35]	-95 [-120, -55]	-145 [-180, -120]	-55 [-95, -30]	-85 [-120, -50]	-130 [-180, -115]	-35 [-73, -15]	-55 [-95, -25]	-120 [-160, -85]
Hypo detection time patient median (min)	-65 [-94, -33]	-90 [-118, -56]	-170 [-180, -124]	-60 [-83, -31]	-80 [-110, -47]	-150 [-180, -123]	-35 [-66, -16]	-55 [-88, -26]	-120 [-166, -101]
BG level at alarm (mg/dL)	58 [55, 60]	64 [62, 66]	78 [74, 79]	55 [53, 57]	62 [60, 64]	75 [71, 77]	52 [48, 54]	55 [51, 60]	72 [61, 75]
false alarms	1 [0, 2]	1 [0, 2]	2 [1, 3]	1 [0, 1]	1 [0, 2]	2 [1, 3]	0 [0, 1]	1 [0, 1]	1 [1, 2]

Despite the sensor noise, there is a clear improvement in the time of detection of hypoglycaemic episodes using CGM sensors compared to standard clinical procedures measuring 1-2 hourly with, in this case, SPRINT. The degree of improvement depends on the design and parameters of the alarm algorithm presented. The reliability of the alarm, as measured by the number of false alarms decreases with increasing advanced detection time.

As a comparison, Table 3 shows the results of using a simple glycemic level alarm algorithm. The alarm is triggered in this case by 3 consecutive CGM readings below the threshold BG value and a negative average gradient over those 3 points. It is obvious that the performance is significantly worse than the integral-based algorithm. False alarm values of -1 indicate that no alarms were triggered at all despite the hypoglycemic event occurring in each case. Thus, the results in this table show that

for more than 75% of the time, with a threshold value of 50 mg/dL, no alarm was triggered despite the oncoming, glycemically near, hypoglycemic event.

Table 3: Early detection of hypoglycaemic episodes reported for the simple glycemic threshold value-based alarm algorithm for all 700 Monte Carlo runs. Data are median [IQR]. A negative value in false alarms indicates that no alarm was triggered.

Consecutive points below alarm threshold	3		
Average gradient at threshold (mg/dL.min)	< 0		
Alarm threshold (mg/dL)	50	60	70
Hypo detection time cohort (min)	-15 [-25, -5]	-40 [-75, -25]	-75 [-110, -45]
Hypo detection time patient median (min)	-10 [-26, -10]	-40 [-69, -25]	-75 [-105, -41]
BG level at alarm (mg/dL)	44 [41, 46]	50 [44, 53]	55 [52, 62]
false alarms	-1 [-1, -1]	-1 [-1, 6]	7 [-1, 13]

Figure 2 shows a typical example of filtered CGM data for one patient run plotted alongside the actual measured 1-2 hourly blood glucose values. The dark band and line show the median (line) and IQR (shaded band) of the CGM hypoglycemic event detection for the entire 100 Monte Carlo runs over this patient, for a window length of 7 samples and integral threshold of 10 mg.min/dL. First, it is evident that even when filtered, the CGM data is still relatively noisy. This noise can cause false alarms and impact on the reliability of the method.

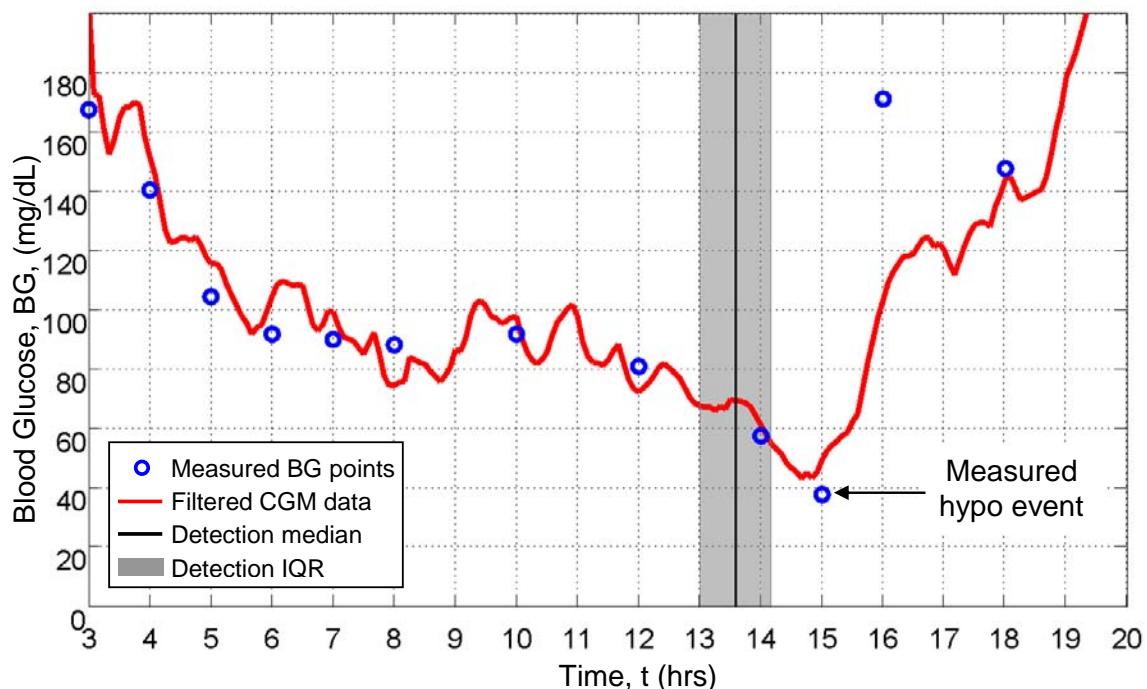


Figure 2: An indication of the early hypoglycaemic detection provided by CGM sensors for one patient showing the median [IQR] for detection over all 700 Monte Carlo runs. The measured data (circles) and one example of the filtered CGM data (line) are shown for context.

Second, the range of hypoglycemia detection is still well in advance of the measured hypoglycemic event at 15 hours. The hypoglycemic value is 39 mg/dL and thus it may also be assumed that it likely occurred very near this time. The IQR of detection is 50-120 minutes before the event, which was preceded by a stable range of glycemia around 90 mg/dL for several hours.

Figure 3 illustrates how false alarms can occur. The integral window length and threshold were 7 samples and 10 mg.min/dL, respectively. There were two false alarms recorded at 14.8 and 17.2 hours. Although the actual blood glucose measurements remain fairly constant, the noisy CGM data pulled the value of the integral below the threshold value, triggering an alarm. The actual alarm wasn't triggered until 21.3 hours, 45 minutes before it was detected by clinical measurements. Hence, in this case, the actual event was detected and reliably alarmed but minor hypoglycemia at 15-18 hours (60-65 mg/dL) triggered false alarms in this particular Monte Carlo run. Other Monte Carlo runs did not trigger false alarms.

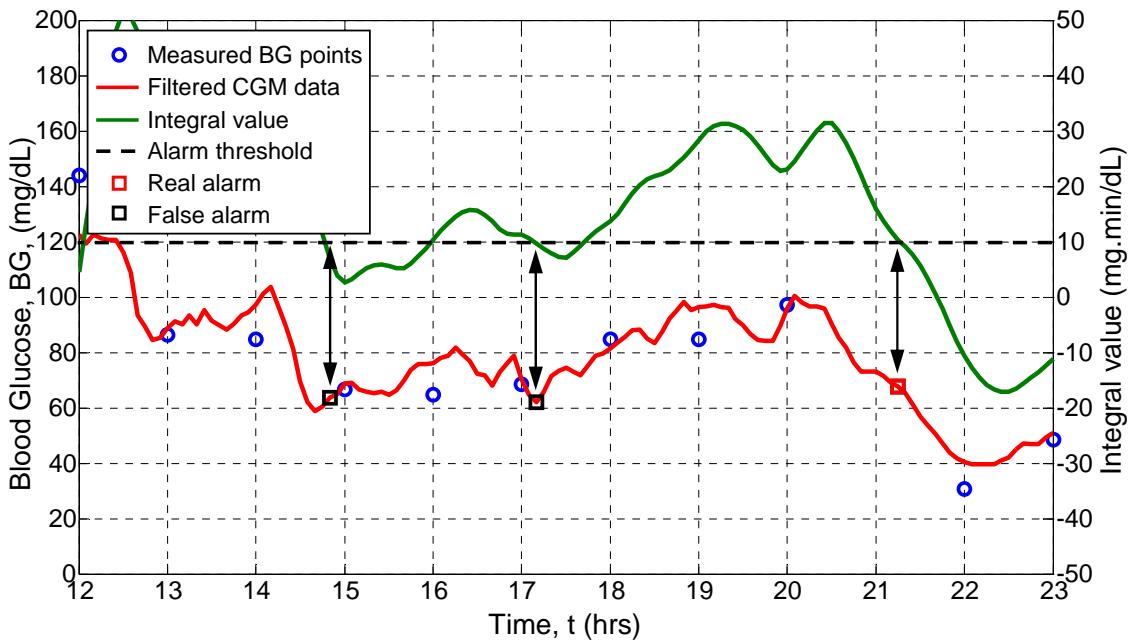


Figure 3: An illustration of false alarms triggered during a simulation

4.0 Discussion:

While CGMs are well studied in ambulatory type 1 diabetics [29, 30], there have been relatively few successful investigations of CGMs in critically ill in-patients [28]. There are several important differences between critically ill patients and ambulatory type 1 diabetics, which may have an impact on the performance of the CGM device. Critically ill patients are generally insulin resistant [1-7] due to high levels of counter regulatory stress hormones and consequently hyperglycaemic without adequate control. Insulin resistance coupled with continuous feeding and sedation, limits the rate of change of blood glucose levels, preventing rapid changes that can occur with meals and exercise and possibly resulting in improved CGM performance compared to ambulatory individuals. On the other hand, peripheral oedema, a lack of dynamic circulation of the interstitial fluid due to the non-ambulatory state [42] may reduce performance.

4.1 Performance:

The results of this study show that using CGM sensors for early hypoglycaemic event detection in an ICU can be very effective. Depending on the design of the filter and alarm algorithm, reliably detecting hypoglycaemic events 1-2 hours before they would normally have been picked up is quite feasible, allowing very early intervention. Early detection and intervention may reduce or eliminate both the events themselves and thus any harm caused by these episodes.

Filtering the raw CGM output with the median filter/LMS fit provides better results than using a more common FIR filter (results not shown). The raw, noisy CGM data had a MAPE of 17.4%, the median/LMS filter reduced this to 9.3% for this cohort. To provide a comparison with “standard” filtering methods, a 9th order low-pass FIR filter designed using the window method results in a MAPE of 13.7% which is much higher than this filtering approach for the same data.

Using an integration-based algorithm to trigger the alarm is much more effective than a simple consecutive value-based or derivative-based methods. The residual noise, post-filtering is still enough to trigger multiple false alarms with a high threshold, or to prevent any alarms at all with a lower threshold when simply using the filtered BG values to trigger the alarm. Table 3 illustrates this point, particularly with an alarm threshold of 60 mg/dL. No alarms were triggered in more than 50% of the simulations with this threshold value, but in 25% of simulations up there were 6 or more false alarms triggered as the IQR 75th percentile is 6 in Table 3 for some cases. Derivative based algorithms were not considered in this study as they are known to be highly susceptible to noise whereas integrals filter noise much like a basic FIR or other digital filter. The integral-based algorithm is thus more robust to noise and therefore provides a much more reliable alarm.

Sensor error due to bias has not been modeled in this research, as it is a calibration issue rather than strictly noise. However if present and positive (BG actually lower than measured), bias would reduce the lead time for hypo detection. If negative bias was present (BG actually higher than measured), the detection lead time would be improved, however the number of false positive detections may also increase. The potential presence of sensor bias therefore necessitates careful calibration of the device and tuning of the detection algorithm to ensure good performance.

Tuning the integral algorithm to optimise the performance involves adjusting the window length and the alarm threshold parameters. There is a trade-off with integral

window length, where long windows provide a more reliable alarm, but with less advance warning than shorter windows. With a long window, the integral takes longer to fall below the threshold once the blood glucose starts dropping. This trade-off also explains why a lower alarm threshold parameter is more reliable, but provides less warning than a higher value.

Selecting the optimal values for window length and alarm threshold would ideally be done over a larger set of patients than used in this proof of concept analysis. This study has a number of limitations however, the method presented can be readily generalised to ICU populations. This performance would justify a pilot clinical trial, not only to validate the detection algorithm, but also the assumptions behind the simulated noise.

4.2 Limitations – Clinical:

The small number of patients in this study (7) may mean the results are not representative of the overall population behaviour. Repeating the study with a larger cohort would provide a more statistically powerful result. However, in the Christchurch ICU all patients receiving insulin are on the SPRINT glycaemic control protocol and therefore very unlikely to suffer a serious hypoglycaemic event (~4% of patients or less), making it difficult to recruit a large cohort. Data from a different study might alleviate this issue for in silico studies, or a pilot clinical trial should be developed.

A further limitation is that only hypoglycaemic episodes at or below 40 mg/dL were investigated in this research. The SPRINT protocol controls patients between 72 mg/dL and 108 mg/dL [11, 25] without significant prejudice towards lower values in this range. Hence, examining a higher hypoglycaemic level (e.g. moderate hypoglycemia < 60 mg/dL) is infeasible as getting closer to the target band results in more false alarms. Uncontrolled or poorly controlled patients may permit the investigation of CGM hypoglycemic detection performance on less severe episodes.

Finally, although the virtual patient simulation method is clinically well validated [25, 34, 43-45], this in-silico study needs to be confirmed with clinical testing. The actual blood-glucose sequences used were model-based, derived from 1-2 hourly clinical measurements with added noise, not real CGM output data. Testing and validation of the findings in a clinical setting, particularly the noise model and will confirm the results.

4.3 Limitations – Signal Processing:

Signal processing and noise model limitations of this study were that we only considered a conservative noise model, simple filtering methods and ignored calibration drift or bias. We assumed non-biased, random independent noise based on reported data.

CGM sensor error consists of a bias due to calibration drift with a random, or quasi-random noise superimposed on top. Calibration drift due to sensor degradation over time was not considered in this study. Without correction, calibration drift will show up as though the actual BG measurements were higher or lower in a relatively consistent manner as the sensor gain drifts [40]. This drift would cause the alarm to trigger early (possibly falsely) or late. However, such calibration drift is very much a function of the frequency and quality of calibration measurements which can likely be controlled more readily in a critical care setting.

The random component of noise modeled in this study was assumed to be independent and normally distributed. The assumption of independent errors may not model the actual CGM sensor noise perfectly, but with no reported time-series data available in literature (to the authors' knowledge), a more comprehensive model was not possible.

The Gaussian noise model employed was conservative ($MAPE = 17.4\%$ for this cohort) as it was based on data from a 2004 study [28] and there have been significant advances in emerging CGM sensor hardware and software since that time. Improved sensor noise characteristics combined with more advanced filtering techniques would result in a much cleaner CGM data stream and hence more reliable alarms. However, the higher noise levels used thus provide a conservative test of the approach.

This study only investigated simple median/LMS and FIR filters. More advanced filtering techniques, such as adaptive filtering and Kalman filtering have been shown to produce very good results [39, 40]. Kalman filtering can also be used to correct for calibration drift [40], which was not studied in this case. However, it is difficult to compare performance here as all have used different data with different noise or error distributions from the sensors.

4.4 Limitations – Analysis Method:

It should be noted that this proof of concept study examined only hypoglycemic events. Thus, there were no opportunities to examine “false positive” alarms. The false alarms here occurred significantly before, or in some cases after, the hypoglycemic event. The false alarms here are thus more representative of a lack of robustness.

However, as seen in Figure 3, moderate hypoglycemia, which is corrected before the severe hypoglycemic event, can still trigger a false alarm if the thresholds are not perfectly optimised – an effort that may be nearly impossible given the variety of patient behaviours that can be encountered. That said, a trigger at this level is still clinically useful and definitely not spurious with regard to treating. Thus, another view of Figure 3 would be that these false alarms, if not frequent may be earlier precursors of hypoglycemic dynamics in the patients response to the intervention, which could potentially be used to moderate care.

4.5 Summary:

Finally, while these are significant limitations, this paper is focused on proof of concept for a novel integration based method of hypoglycemia detection. The primary goal was to develop a method for accurate, reliable alarms with good warning that account for realistic or greater sensor noise that was also readily generalisable to similar sensors. Thus, the larger noise used is a conservative test and application to cases with lesser noise will provide better results. The main comparison to other alarm methods should then be in terms of relative performance presented here. Overall, the good result with sub-optimal filtering and a conservative noise model only serves to show that the method is feasible and deserves further investigation.

Finally, examining the SPRINT and prior pilot trial results, hypoglycemia could have been avoided without using CGM if measurements were made every 30 minutes (48/day). However, this rate potentially is invasive to the patient and far too burdensome for regular practice in a clinical ICU environment. Hence, CGM offers the opportunity to observe trends semi-invasively and reduce clinical burden, while also increasing potential safety by mitigating the tradeoff between the burden of more frequent measurement, and the safety and tight glycemic control outcomes that it can provide. In SPRINT, even 1-2 hourly measurements yielded only 10 events over 8 patients (2%), a majority of which were due to clinical errors. So, the methods presented here are thus about reducing a potentially small number of events to zero as well as mitigating the clinical burden of frequent measurement.

5.0 Conclusions:

This paper has developed and analysed, *in silico*, a hypoglycemia alarm detection method for application with CGM sensors in critical care or, potentially, other hospital patients. While the results remain to be verified in a pilot clinical trial, the overall method appears robust and accurate to conservatively large sensor noise. The main conclusions and outcomes drawn include:

- A median filter with LMS smoothing was presented and shown to be robust at removing significant sensor noise with minimal lag compared to standard FIR filters and similar to reported results for more advanced filters.
- An integral based method was developed and seen to provide robust hypoglycemia detection with significant lead time before the event to enable intervention. The number of false alarms was very low and performance was better than the simple use of thresholds by using the integral approach.
- The results presented justify a pilot clinical trial of the methods.

Finally, the methods presented are readily generalisable to other hypoglycemic levels or control protocols.

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