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Abstract

Auditory alarms communicate critical patient information to clinicians, yet poor alarm design contributes to alarm fatigue, which can result in delayed clinician responses and compromised patient safety. This study evaluated novel auditory heart rate (HR) and blood pressure (BP) alarms that were designed using perceptually informed principles to improve alarm identification. We tested these novel alarms in realistic co-alarming conditions to better understand how proprietary General Electric (GE) and Philips alarms affect identification of the novel alarms. Thirty-nine adult participants completed a laboratory-based experiment in which the novel HR and BP alarms were presented in isolated and co-alarming conditions. Participants identified alarms they heard from choices shown on a computer screen and were evaluated for their response times and accuracy. Responses were 763 ms slower to HR alarms than BP alarms. The presence of proprietary alarms during co-alarming produced an 89 ms increase to reaction time and a 4% decrease in accuracy while identifying novel alarms. Compared to trials with no co-alarming, responses to ascending HR alarms were 183 ms slower during co-alarming and responses to descending HR alarms were 4 ms slower during co-alarming. Practically, these differences in reaction time and accuracy may influence clinical decision-making despite being brief relative to clinical response windows. These findings highlight the importance of temporal and spectral structure in auditory alarm design and suggest that carefully designed alarms can maintain perceptual clarity in complex acoustic environments to be tested and implemented.

Key Words: Clinician performance | Multisensory alarms | Patient safety | Auditory icons

Introduction

Alarm Fatigue and Clinical Consequences

Auditory alarms are essential for signaling clinically significant changes in patient status, but their growing number and poor perceptual design have contributed to alarm fatigue. Alarm fatigue occurs when clinicians become desensitized to frequent, non-actionable, or indistinct alarms, resulting in delayed responses, missed alerts, and compromised patient safety [1]. The Joint Commission designated alarm safety as a National Patient Safety Goal in 2013 [2]. In high-acuity environments such as intensive care units and operating rooms, clinicians may encounter up to 700 alarms per patient per day, with nearly 80-99% being false or clinically insignificant [3]. This high volume increases cognitive load and contributes to clinician stress and burnout. It also negatively affects patients, contributing to psychological stress and prolonged recovery [4].

Limitations of Current Clinical Alarm Systems

Despite increased attention to alarm management, many monitoring systems still rely on alarm sounds that are difficult to distinguish from one another. Proprietary monitoring systems from General Electric (GE) and Philips often use alarms with similar melodic structures and limited perceptual distinctiveness, making it difficult for clinicians to identify signals when alarms occur simultaneously. Alarms can interfere with one another through perceptual masking, further complicating alarm recognition and interpretation [5].

Vital signs alarms present perceptual challenges because they must convey the identity and direction of physiological change, such as an increase or decrease in heart rate (HR) or blood pressure (BP). In many existing alarm designs, these cues are not present or emerge later in the signal. These limitations present an opportunity to communicate patient information more effectively.

Alarm Design Principles: Acoustic Foundations

Medical alarm acoustics are guided by standards such as the International Electrotechnical Commission's (IEC) collateral standard 60601-1-8 [6]. However, these standards provide limited information on how alarms are perceptually interpreted in complex clinical environments. Previous research has shown that acoustic features such as fundamental frequency, harmonic content, temporal structure, and melodic contour can influence perceived urgency and detectability, offering further guidance for effective alarm design [7].

The melodic direction of an alarm can provide critical information about patient emergencies. For example, an ascending pitch can indicate exceeding normal physiological state, such as tachycardia. In this way, the alarm can indicate both the occurrence and direction of the abnormal condition. Alarms with such distinctive spectral and temporal features may be more resistant to masking from concurrent sounds, as masking can impair alarm recognition [8]. In April 2022,

Amendment 2 to IEC 60601-1-8 introduced auditory icons as tools to improve alarms in six categories of clinical emergencies, including cardiovascular, artificial perfusion, ventilation, oxygenation, temperature/energy delivery, and equipment or supply failure [6]. Auditory icons are sounds that intuitively relate to the physiological event they represent. Bingham et al. (2023) built on these principles by demonstrating that modifying the harmonic structure of BP alarms improved their detectability under challenging listening conditions, providing a foundation for the HR alarm designs evaluated in this study [9].

Auditory Processing and Neural Encoding in Alarm Perception

Auditory perception relies on neural processes that transform acoustic signals into perceptual categories. During the cortical integration of sound, temporal and spectral information are jointly analyzed, which includes salient perceptual features such as pitch, timbre, and rhythmic envelope [10]. When multiple sounds share overlapping spectra or similar melodic contours, segregation becomes inefficient, resulting in delayed recognition [11, 12].

Study Objectives

This study evaluated the effectiveness of novel auditory HR and BP alarms designed using perceptually informed acoustic principles to improve detectability and interpretability in complex alarm environments. We assessed response time and identification accuracy for these alarms in isolated and realistic co-alarming conditions involving proprietary GE and Philips alarms. We included masker alarms to better understand the effects of competing alarms on alarm interpretation and determine whether our novel alarms are ready for clinical testing. Our novel alarms would not replace current monitors using proprietary alarms that would remain active to signal ongoing patient emergencies during implementation.

We hypothesized that the presence of maskers will increase response time and decrease accuracy when identifying the novel alarms. In addition, we conducted exploratory analyses to compare response time and accuracy for the novel alarms as a function of alarm type and alarm direction when presented alone and with maskers. We did not predict differences in responses to the novel HR and BP alarms, given the similarities in their design principles.

Methods

Participants

Thirty-nine healthy adult students at Vanderbilt University were recruited for this study. Participants were required to be at least 18 years old, proficient in English, and free from known hearing impairments. Screening was completed by participant self-report. Individuals reporting hearing abnormalities were excluded. Recruitment was carried out via flyers posted in academic buildings and students were offered course credit for participating in this study. Informed consent was obtained electronically using REDCap. All study procedures were approved by the Vanderbilt University Institutional Review Board (IRB #242060, approval date: 01/27/2025).

Of the 39 enrolled participants, all were retained for final analysis. All participants complied with the experimental tasks and no technical issues resulted in incomplete or unreliable data.

Alarm Design

The HR alarms were newly developed for this study and underwent their initial evaluation. Each alarm consisted of three sequential auditory components: a cardiovascular “lub-dub” auditory icon, a high-priority urgency pointer, and a directional HR cue. These components were consistent with the IEC’s guidance for high-priority alarms [6]. Each component was one second in duration, producing a three-second alarm presentation.

The IEC’s recent guidance introduced auditory icons, which are sounds modeled after real-world cues, such as a “lub-dub” heartbeat representing cardiovascular alarms. The cardiovascular component established the physiological identity of the signal. It consisted of two pulse events approximating systolic and diastolic phases.

The IEC standard also incorporates pointers that communicate acuity levels through pulse patterns, with a rapid five-pulse sequence indicating high urgency. A high-priority pointer was implemented as a sequence of brief tone bursts centered on A4 (440 Hz) [6]. Each pulse had a short duration with brief silent gaps between pulses, forming a five-pulse pattern intended to communicate urgency.

The HR directional cues used dissonant intervals to be perceptually associated with physiological deviation. Bradycardia alarms used a downward-perceived pitch motion built around a tritone, emphasizing greater low-frequency energy and a heavier bass presence. Tachycardia alarms used an upward-perceived pitch motion built through an inverted tritone, with stronger mid-high frequency content. These contrasting structures were selected because the tritone produces an unstable and attention-capturing quality, and the inversion supported a clear distinction between bradycardia and tachycardia.

The resulting bradycardia (Online Resource 1)¹ and tachycardia (Online Resource 2)² alarms combined each of these components to reinforce the semiotic mapping between sound structure and physiological meaning.

The BP alarms used in this study were adapted from our lab's previous work [9]. A hypotension alarm (Online Resource 3)³ with a downward-perceived pitch motion and a hypertension alarm (Online Resource 4)⁴ with an upward-perceived pitch motion were adapted from this study. The BP alarms contained the same high-priority urgency pointers as the HR alarms, but unique BP auditory icons and directional cues. The BP alarms served as validated reference stimuli against which the HR alarms were evaluated.

Experimental Design

This study employed a within-subjects design to evaluate participants' ability to identify auditory physiological alarms under cognitive load. Each participant completed a 90-minute session in a quiet laboratory room, a controlled auditory environment. Participants were seated at a desk in front of a computer display presenting the experimental interface, with the experimenter seated nearby to monitor. The study used a 2 x 2 x 5 design. The first factor, alarm type, had two levels: HR and BP. The second factor, alarm directionality, had two levels: ascending and descending. The third factor, masker, had five levels: no masker, GE Yellow, GE Red, Philips Yellow, and Philips Red.

Participant Orientation

Upon providing informed consent, participants completed an orientation to the auditory alarms and experimental interface. They were introduced to four distinct alarms: two novel alarms representing high and low HR conditions and two alarms representing high and low BP conditions.

Participants then completed a training assessment in which they listened to each alarm and identified which physiological condition it represented. Participants were required to score 100% before progressing to the experimental testing phase. The training assessment could be repeated to meet this criterion.

Masker Design

Each alarm was tested under five masking conditions: no masking (alarm presented alone) and masking with the proprietary medium and high-acuity alarms from General Electric HealthCare (GE, Chicago, IL) and Philips North America (Philips, Cambridge, MA) (four combinations of

¹ Novel bradycardia alarm, including cardiovascular auditory icon, high-priority pattern, and downward-perceived pitch motion.

² Novel tachycardia alarm, including cardiovascular auditory icon, high-priority pattern, and upward-perceived pitch motion.

³ Hypotension alarm [9].

⁴ Hypertension alarm [9].

acuity and manufacturer). The GE High Priority Physiological Alarm (Red) (Online Resource 5)⁵ and GE Medium Priority Physiological Alarm (Yellow) (Online Resource 6),⁶ which are utilized as auditory alarms in GE patient monitors such as the CARESCAPE™ R860, were used in this study [13]. The Philips High Priority Patient Alarm (Red) (Online Resource 7)⁷ and Philips Medium Priority Patient Alarm (Yellow) (Online Resource 8),⁸ which are utilized in Philips monitors such as the IntelliVue Patient Monitor MX100/X3, were used in this study [14]. The masking alarms were recorded from clinical monitors and digitized for use.

Dual-Task Cognitive Load

To simulate the multitasking demands and cognitive load in clinical environments, participants simultaneously completed a visual discrimination task throughout the alarm trials, following a dual-task paradigm previously validated in alarm perception research [15]. In this task, red and white letters flashed sequentially on the screen every 333 ms. Participants were instructed to press the computer spacebar whenever a red vowel appeared.

Experimental Testing Phase Structure

The experiment consisted of eight blocks, each containing 18 randomized alarm presentations for a total of 144 trials across the session. During each presentation, the alarm played twice with a one-second silent interval between repetitions. After the second repetition, a silent interval that ranged randomly from 8 to 23 seconds occurred before the next alarm began. Alarm cycles, including alarm presentation and silent intervals, averaged 22.5 seconds in length (Fig. 1a). Participants could identify the alarm condition anytime within the alarm cycle. Two-minute breaks were provided between blocks, totaling approximately 14 minutes of rest across the session.

Participants wore Sennheiser HD 280 Pro headphones, calibrated to maintain a listening volume between 60 and 70 decibels in accordance with American Speech-Language-Hearing Association safety guidelines [16]. Participants were instructed to identify the physiological state by clicking one of four labeled options on the computer screen: low BP, high BP, low HR, or high HR (Fig. 1b). The alarm was terminated once a selection was made. Response time was measured from alarm onset to the participant's click, and responses were scored based on whether the correct condition was selected. Alarm accuracy was calculated as the number of correctly identified alarms divided by the total alarm presentations for each condition. Trials in which participants selected an incorrect alarm category were excluded from response time analyses but retained for accuracy calculations. This approach ensured that response time measures reflected correct alarm identification, while accuracy metrics captured overall performance including false positive and false negative responses.

⁵ High-pitch burst.

⁶ Moderate-pitch burst (compared to GE Red).

⁷ High-pitch burst.

⁸ Moderate-pitch burst (compared to Philips Red).

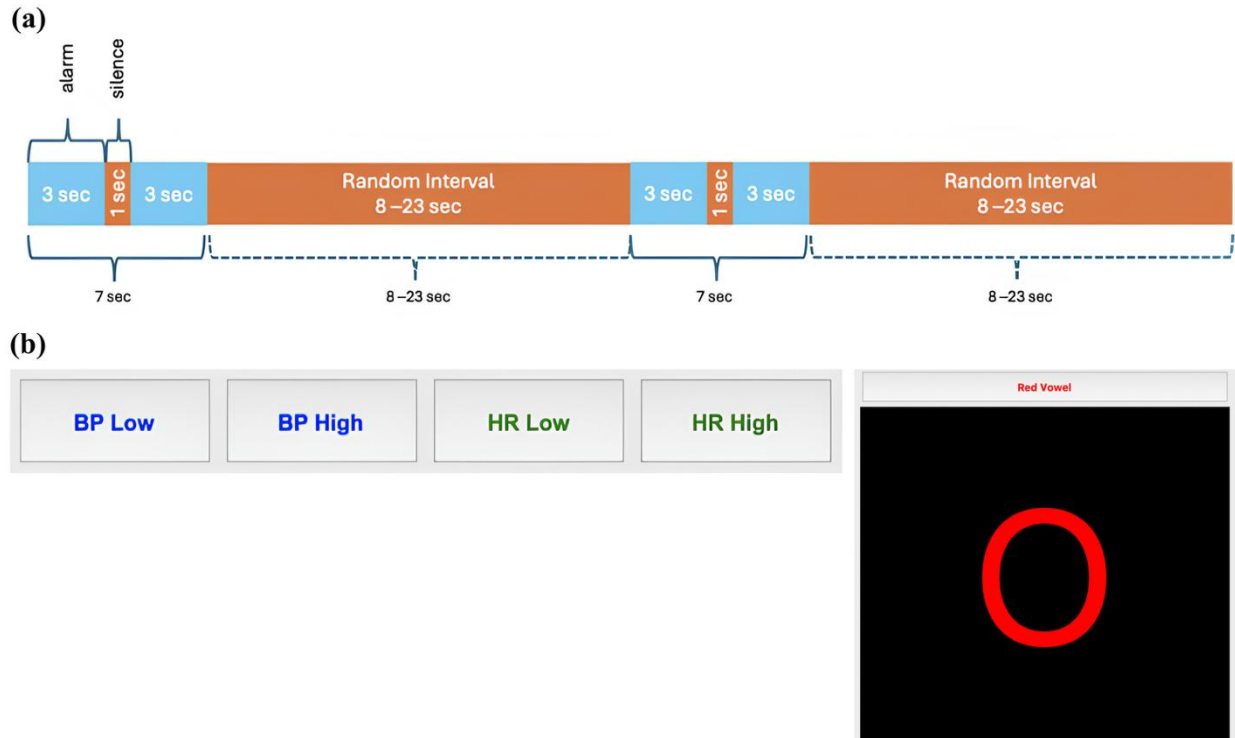


Fig. 1 (a) Illustration of the experimental timing structure **(b)** Illustration of the computer interface participants interacted with, including buttons to identify sounds and complete the visual discrimination task

Statistical Methods

Analysis of Response Time and Accuracy at Baseline

We compared baseline identification of the alarms with no auditory maskers. To examine reaction times to the alarms, we performed a two-way repeated-measures ANOVA, using alarm type (HR or BP) and alarm direction (ascending or descending) as factors and mean response time for the conditions where no masker was present as the dependent variable.

We also compared baseline accuracy for the alarms with no auditory maskers present. A two-way repeated-measures ANOVA using response accuracy as the dependent variable and alarm type and alarm direction as factors was performed.

Analysis of Auditory Maskers on Alarm Identification

To compare alarm identification with and without auditory maskers, we calculated a mean response time for correct responses and mean accuracy for each trial where a masker was present, regardless of the masker's identity. Then we compared these values with the baseline mean response times accuracy values.

We performed a three-way ANOVA using mean response time as the dependent variable and alarm type, alarm direction, and masker presence as factors. Next, we performed a three-way ANOVA using response accuracy as the dependent variable and alarm type, alarm direction, and masker presence as factors.

Analysis of Alarm Direction Under Masking

We also examined the effects of masker characteristics on response time for each of the two HR alarm directions. We calculated four mean response time difference scores for each HR alarm by taking the difference between mean response time to the HR alarm when each of the four maskers was present and the mean baseline response time for that HR alarm. Thus, we obtained eight reaction time difference scores. We performed a three-way ANOVA using mean reaction time difference as the dependent variable and alarm direction, masker manufacturer (GE or Philips), and masker priority (Red or Yellow).

Similarly, we calculated mean difference scores for accuracy. For the two HR alarms, we calculated the difference in mean accuracy when each of the four maskers was present and when no masker was present, resulting in eight mean accuracy difference scores. We performed a three-way ANOVA with mean accuracy difference as the dependent variable and alarm direction, masker manufacturer, and masker priority as factors.

Data Cleaning

In reviewing the response time data, six participants had at least one masking condition where they did not make any correct responses, resulting in a missing value. A total of 26 mean response time values were missing across the 39 participants (3.33% of total values). To apply standard repeated-measures ANOVA analyses, we replaced these missing values using a by-item, by-subject replacement method [17], where the missing value was replaced using the weighted mean of the existing mean response time values for the subject and the existing mean response time values for the item. No accuracy data were missing, as the accuracy for each condition for each participant was calculated as the proportion correct.

The distribution of response times was positively skewed. All response time analyses were run using both raw reaction times and $\log_{10}$ -transformed reaction times as the dependent variable. With one exception, the results were identical whether raw reaction times or $\log_{10}$ -transformed reaction times were used as the DV. For ease of interpretation, we report the results of the analyses using raw reaction times, except in the place where $\log_{10}$ -transformed reaction times produced a significant result that did not occur with raw reaction times.

Results

Analysis of Response Time and Accuracy at Baseline

This analysis found a main effect of alarm type ($F(1,38) = 97.58, p < 0.001, \eta^2_p = 0.72$). Responses to HR alarms ($M = 3900$ ms, 95% C.I. [3789, 4011]) were slower than responses to BP alarms ($M = 3137$ ms, 95% C.I. [2929, 3345]) (Fig. 2).

We did not find any main effects or interactions for baseline accuracy in response to the alarms with no maskers present.

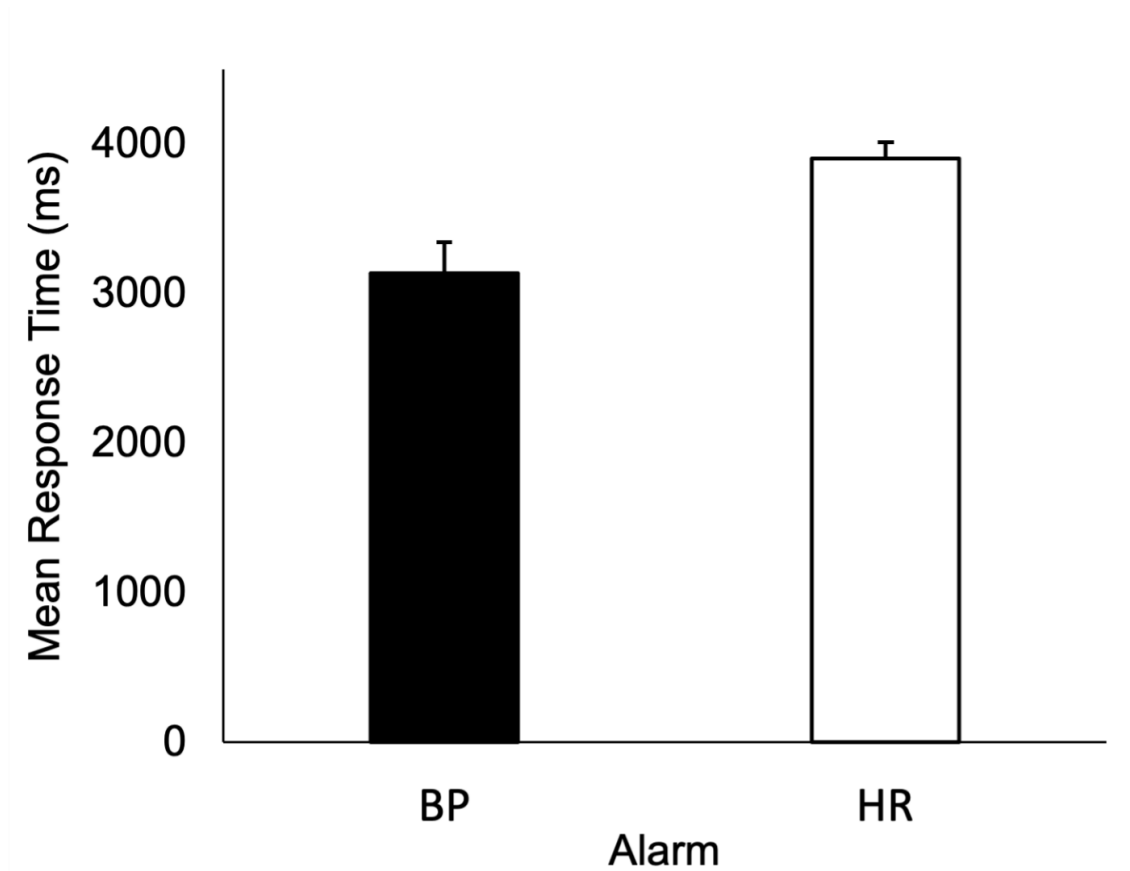


Fig. 2 Mean response time to novel alarms with no maskers present. Participants responded more quickly to BP alarms than HR alarms. Error bars represent the 95% confidence interval for the mean

Effect of Auditory Maskers on Alarm Identification

There was a main effect of alarm type ($F(1,38) = 98.95, p < 0.001, \eta^2_p = 0.72$). Responses to HR alarms ($M = 3947$ ms, 95% C.I. [3831, 4063]) were slower than responses to BP alarms ($M = 3179$ ms, 95% C.I. [2961, 3398]). Second, there was a main effect of masker ($F(1,38) = 6.58, p < 0.05$,

$\eta^2_p = 0.15$). Responses to alarms were slower when a masker was present ($M = 3607$ ms, 95% C.I. [3435, 3779]) than when no masker was present ($M = 3518$ ms, 95% C.I. [3371, 3666]) (Fig. 3a). Using $\log_{10}$ -transformed reaction times as the DV also revealed an interaction of alarm direction and masker ($F(1,38) = 6.63$, $p < 0.05$, $\eta^2_p = 0.15$). Figure 4 shows this interaction. The presence of the masker slowed response time for ascending alarms but not for descending alarms.

There was a main effect of masker presence ($F(1,38) = 16.49$, $p < 0.001$, $\eta^2_p = 0.30$) (Fig. 3b). Response accuracy was lower when a masker was present ($M = 0.87$, 95% C.I. [0.82, 0.92]) than when no masker was present ($M = 0.91$, 95% C.I. [0.87, 0.95]).

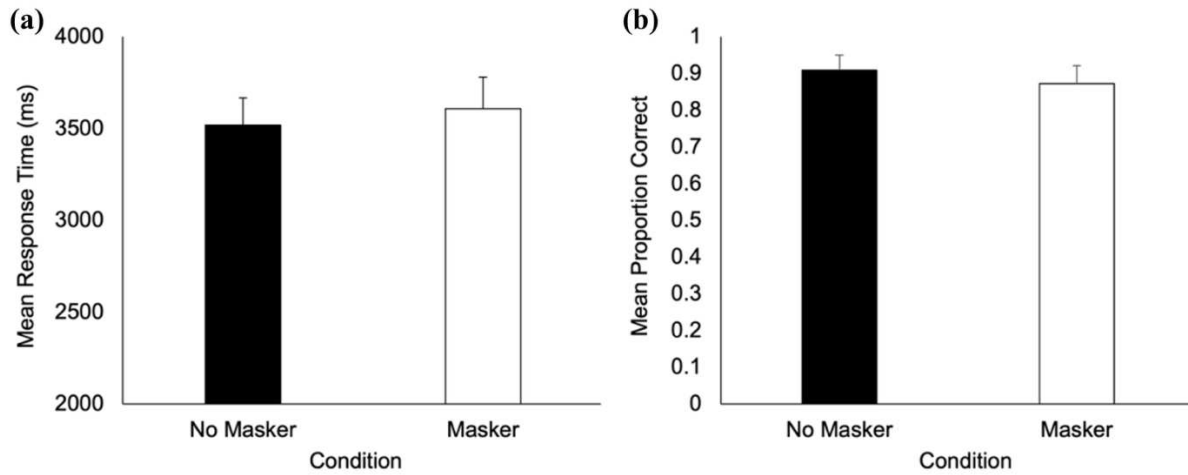


Fig. 3 Effects of a masker on (a) mean alarm identification response time and (b) mean alarm identification accuracy. Error bars represent the 95% confidence interval for the mean

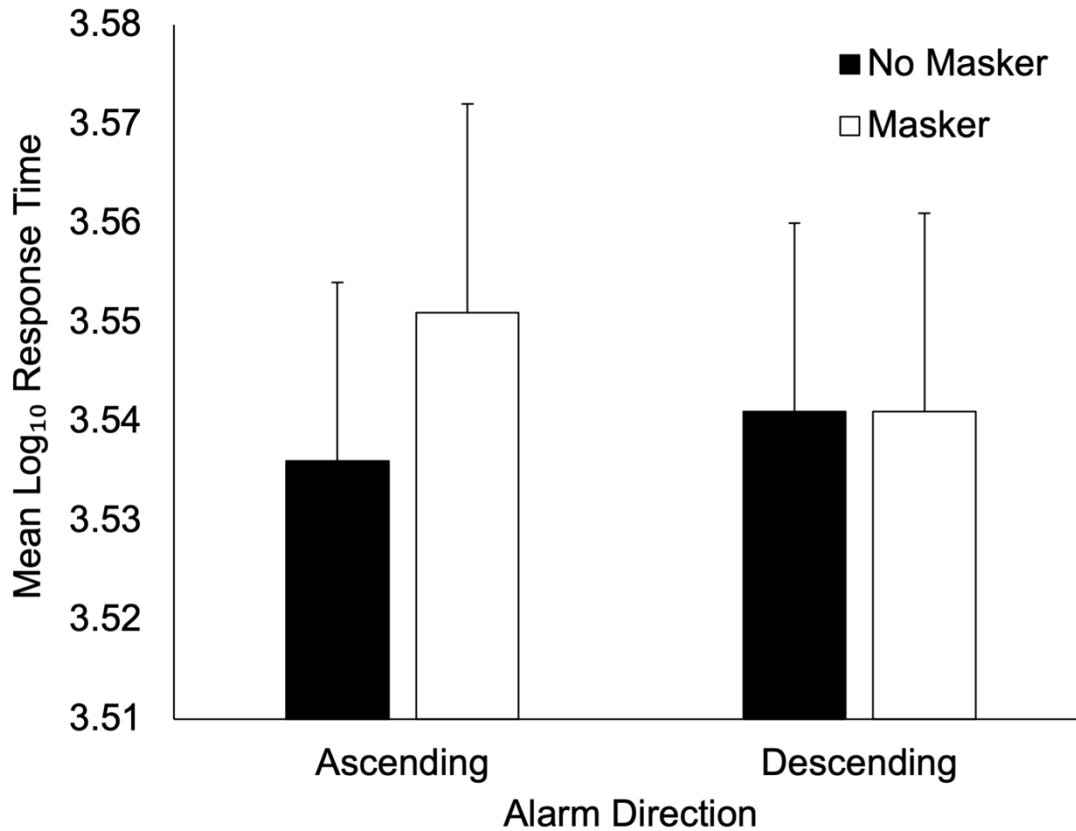


Fig. 4 Interaction of masker and alarm direction on mean log₁₀ response times. Error bars represent the 95% confidence interval for the mean

Effect of Alarm Direction Under Masking

There was a main effect of alarm direction ($F(1,38) = 4.83, p < 0.05, \eta^2_p = 0.11$). Ascending alarms were 183 ms slower when a masker was present (95% C.I. [70, 297]) than when no masker was present, whereas descending alarms were only 4 ms slower when a masker was present, compared with when no masker was present (95% C.I. [-109, 117]) (Fig. 5).

There were no main effects or interactions for the effect of alarm direction under masking on accuracy.

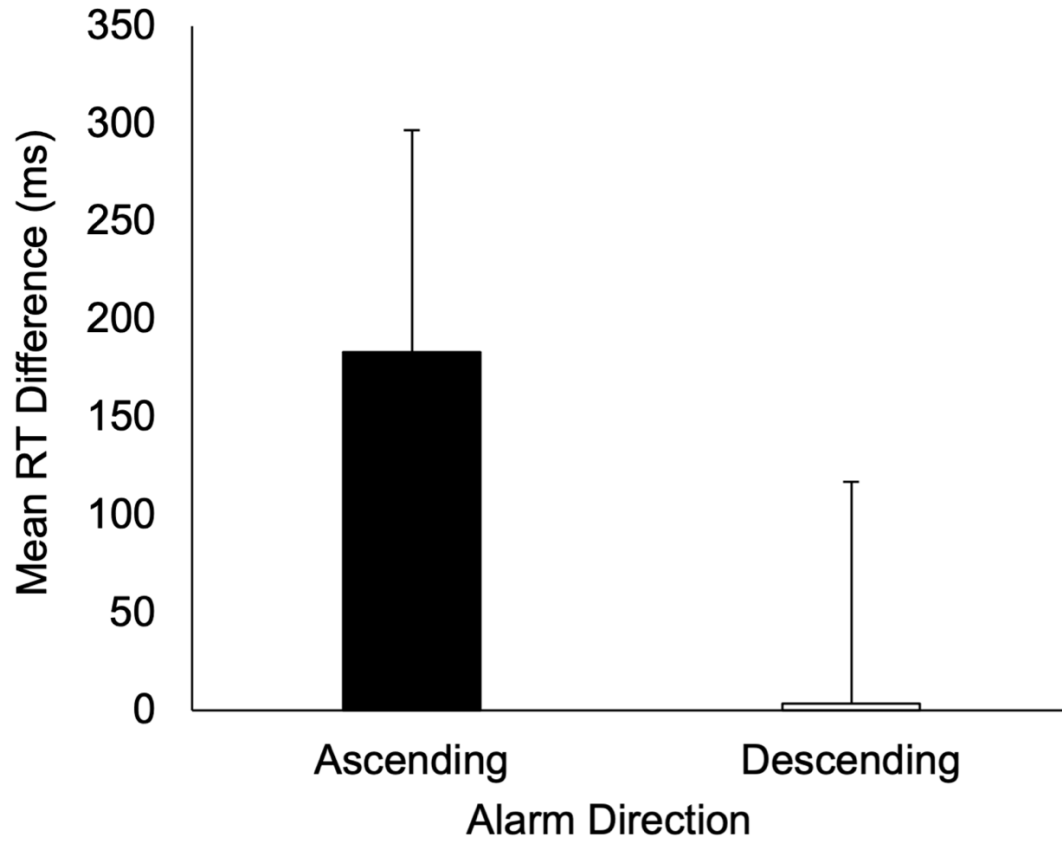


Fig. 5 Effect of alarm direction on mean response time difference. Error bars represent the 95% confidence interval for the mean

Discussion

Three key points emerge from this study. First, the novel HR alarms we developed elicited slower responses compared to the BP alarms, whether the alarms were presented in isolation or with a masker. Second, the presence of a masker increased response time and decreased response accuracy. Third, in co-alarming conditions, ascending alarms received slower responses than descending alarms.

Alarm-Type Effect

Correct alarm identification was slower for HR alarms than for BP alarms. This difference suggests the temporal distribution of acoustic information affects alarm recognition. The BP alarms' front-loaded structure provided earlier perceptual cues. In contrast, the HR alarms required listening for a longer duration before the defining contour signaled the physiological direction of change. This delay likely shifted the moment of categorical decision-making later, resulting in slower response times.

These findings align with prior work showing that alarms which concentrate discriminative features near onset enhance detectability and reduce masking [9, 12]. Schlittenlacher and Ellermeier (2015) show that sub-second differences in a sound's acoustic onset affect reaction time. Steepening a tone's rise time shortened response time by several milliseconds, with intensity delivered closer to onset strongly influencing the response [18]. Redesigning alarms with more front-loaded cues could help align temporal dynamics with human auditory processing, thereby reducing cognitive effort and improving reaction times to patient emergencies.

Alarm Direction

Participants responded more slowly to tachycardic versus bradycardic HR alarms in the presence of a masker. This suggests the direction of glissando in an auditory alarm affects interpretability. While more acoustically complex alarms are often more resistant to masking, excessive acoustic density in noisy environments can increase cognitive load and hinder interpretation [19]. Slower response times to tachycardic HR alarms may be due to their upward glissando, which overlaps with the higher-frequency range of common clinical alarms such as Philips high-acuity alarms [20]. Bradycardic HR alarms occupy less crowded frequency regions, enhancing their perceptual salience.

According to spectral analysis, our bradycardic alarms demonstrate greater spectral levels below 300 Hz than above 1 kHz, indicating a relative emphasis on lower-frequency content (Fig. 6). Our tachycardic alarms demonstrate more prominent spectral features in the mid-high frequency range, including local maxima at 1-1.5 kHz and 2.5-3 kHz (Fig. 7). Low-mid frequency focused alarms may offer perceptual advantages in dense auditory environments such as ICUs and operating rooms.

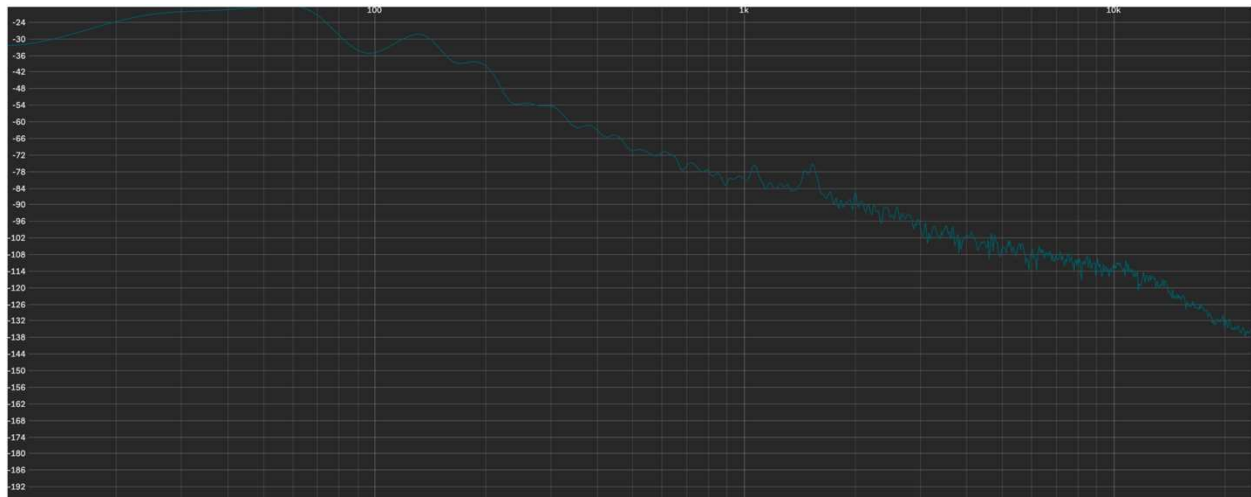


Fig. 6 Spectral analysis of the descending HR alarm (bradycardia). The x-axis shows frequency (Hz) and the y-axis shows amplitude in decibels (dB)

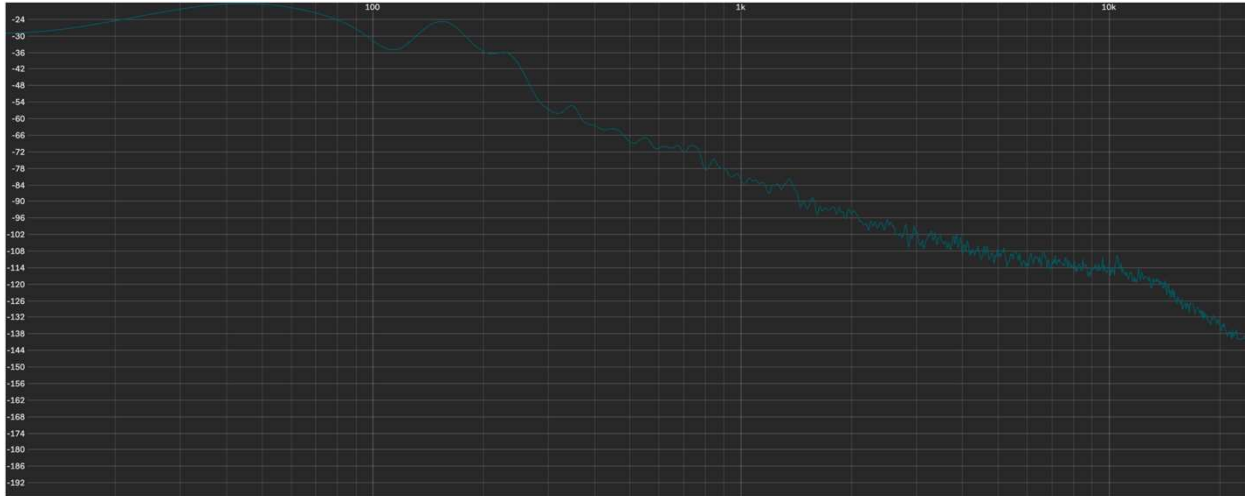


Fig. 7 Spectral analysis of the ascending HR alarm (tachycardia). The x-axis shows frequency (Hz) and the y-axis shows amplitude in decibels (dB)

Effects of Masker Alarms

Responses were 89 ms slower and 4% less accurate when proprietary GE and Philips alarms were present as co-alarms to our BP and HR alarms than when no maskers were present. These results indicate that the presence of competing alarms occludes comprehension of the clinical events presented by the novel alarms. Still, relative to the duration of the alarm and response window, the increase in response time and decrease in accuracy are small, reliable effects that may not result in negative outcomes. Slower clinician response or alarm misidentification may cause delayed, inaccurate action by clinicians on the order of many presentations of our novel alarms alongside maskers. Despite possibility for improvement, our novel alarms should be further tested in clinical settings. Future work should advance our alarm design to minimize the effects of the maskers.

There may be various reasons why these differences were observed. The effect of the masker alarms may be influenced by the signal-to-noise ratio between them and our novel alarms. In this study, alarms were normalized to have the same volume to focus on the effects of other acoustic properties. Current clinical research and practices suggest the use of adaptive volume control and priority-based volume settings to differentiate between alarms [21, 22]. Future work should explore these possibilities.

Limitations and Strengths

Our study contains limitations. The dual-task paradigm we used was a visual letter discrimination task, which does not fully reflect challenges faced by clinicians during patient emergencies, such as conducting medical procedures. However, such tasks have been used to simulate cognitive load previously in similar studies [15]. Future studies should incorporate dual-task paradigms that more closely mimic clinical activity.

Additionally, our study population, for practical reasons, involved medically naive undergraduate participants rather than clinicians. This may limit the generalizability of our findings due to clinicians' experience with alarms such as the proprietary alarms in this study. Future evaluation of our novel alarms will involve clinicians. However, like previous research on clinical alarm development, it was important to test the novel alarms in a laboratory setting with lay participants [12]. Furthermore, during recruitment, participants' hearing abilities were screened using self-report. Future studies should utilize audiometry to assess participants more reliably.

Despite these limitations, this study presents strong novel alarm designs rooted in acoustic principles. The HR alarms were designed using perceptual features including directional pitch cues, harmonic structure, rhythmic organization, and auditory icons. By grounding the alarm design in established principles of auditory perception and human factors research, this study contributes a theoretically informed framework for developing clinical alarms.

Testing these alarms in a controlled laboratory setting offers greater reliability that can safely inform future testing in clinical settings. This environment minimized extraneous variability, allowing differences in alarm identification to be attributed more confidently to the acoustic properties of the alarms. Incorporating a validated dual-task paradigm and proprietary masking alarms further increased the external validity of the experiment.

Future Work

We aim to develop and test our novel auditory alarms in clinical settings. Clinicians are often tasked with integrating various streams of information such as through monitors that present numerical readings alongside auditory alarms. Upon improving the performance of our novel HR alarms, we plan to integrate them with our previously tested visual and tactile alarms [23, 24]. We aim to design a multisensory alarm system that addresses current limitations of hospital alarms.

Conclusion

This study evaluated the performance of novel auditory HR alarms in the presence of competing masker alarms. Participants identified the HR alarms with high accuracy. The presence of concurrent proprietary alarms delayed response times and decreased accuracy of recognition. These findings indicate that the novel HR alarms should be improved to be perceptually distinguishable even under co-alarming conditions, which is a key concern in clinical environments where alarms sound simultaneously. Differences in response time across alarm types and directional contours highlight the role of temporal and spectral structure in shaping alarm perception. Together, these results support the use of perceptually informed acoustic design principles for developing clinical alarms that improve informational clarity in complex auditory settings.

Statements and Declarations

Author Contributions

Shayaan Essani: Writing – review & editing, Writing – original draft, Methodology, Software, Conceptualization, Data curation, Validation, Formal analysis. **Abtin Ghelmensaraei:** Writing – original draft, Methodology, Data curation. **Madison Ferguson:** Writing – original draft, Methodology, Data curation. **Joshua Shive:** Writing – review & editing, Writing – original draft, Validation, Methodology, Formal analysis, Supervision. **Richard H. Epstein:** Writing – review & editing, Investigation, Methodology, Software, Formal analysis, Project administration, Supervision. **Joseph Schlesinger:** Writing – review & editing, Investigation, Methodology, Formal analysis, Project administration, Supervision. All authors reviewed and approved the final manuscript.

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Competing Interests

The authors have no relevant financial or non-financial competing interests to disclose.

Data, Material, and Code Availability

The lab measurements, materials, and code used to in this study are available from the corresponding author upon reasonable request.

Ethics Approval

All study procedures were approved by the Vanderbilt University Institutional Review Board (IRB #242060, approval date: 01/27/2025).

Consent to Participate

All participants provided informed consent electronically through REDCap.

Consent to Publish

Not applicable. The data presented in this study are anonymized and do not contain identifiable information.

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