Title
Determining the Ideal Electrode Configuration for Continuous In-Hospital ECG Monitoring

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Publication Date
2014

Peer reviewed|Thesis/dissertation

Determining the Ideal Electrode Configuration for Continuous In-Hospital ECG Monitoring

By

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DISSERTATION

Submitted in partial satisfaction of the requirements for the degree

of

DOCTOR OF PHILOSOPHY

In

Nursing

In the

GRADUATE DIVISION

Of the

UNIVERSITY OF CALIFORNIA, SAN FRANCISCO

[redacted]

Dedication and Acknowledgements

I dedicate this dissertation to my father, God rest his soul, who supported and inspired me to have the ambition and desire to succeed in making a better life. After many hospitalizations, my father told me that if I wanted to do something meaningful in the hospital, “make these heart monitors better”.

I also must acknowledge my mentor, Dr. Barbara Drew, for her inspiration and encouragement to do something meaningful with my career. For many years, I said that I would never get a PhD, especially in Nursing. Dr. Drew showed me that nurses make meaningful contributions to the science of patient care, not necessarily nursing science or medical science. To Dr. Claire Sommargren, who has continued to push me since day one to get through this program and stay on track, when many distractions were constantly presented to me. And Dr. Hu, I have the utmost of respect for an engineer joining the faculty in a school of nursing. Your and your team made my data analysis possible, and without you, I would not be finishing this research.

My deepest appreciation to the family, friends, and co-workers who picked up my slack when I was too busy to even recognize that I was overloaded with a full-time career and full-time doctoral program. I could never have done this alone.

Abstract

Determining the Ideal Electrode Configuration for Continuous In-Hospital Electrocardiographic (ECG) Monitoring

Richard L. Fidler, PhD(c), MSN, MBA, CRNA, ANP

Significance: Hospital ECG-monitoring is done using the Mason-Likar electrode configuration with chest-mounted and newer technology allows the addition of precordial electrodes to the bedside monitor to acquire a 12-lead ECG. Mason-Likar limb electrodes need to move to the limbs for a standard 12-lead ECG; however, if this step is missed nonstandard and nonequivalent ECG is obtained. The Lund electrode configuration, with more distal limb electrodes was proposed as a solution, but it is unknown how Lund and Mason-Likar compare in signal quality, false lethal arrhythmia alarms, and patient comfort.

Methods: One hundred patients from ICU and PCU were enrolled, and in addition to standard hospital monitoring equipment, each subject wore two Holter monitors, one in the Mason-Likar and the other in the Lund electrode configurations for a 24-hour period. Randomization to abrasive skin prep was conducted. ECG signals were sent for blinded analysis for signal quality using the Hook-Up Advisor® and arrhythmia analysis using EK-Pro®. Signal quality was rated as “green-yellow-red”, and lethal arrhythmia alarms were categorized as true or false by clinicians. Qualitative patient data regarding the monitoring experience was also gathered.

Results: Subjects each provided a mean of 23.8-hours of data in both electrode configurations, and 45 subjects received abrasive skin preparation. Signal quality was compared between configurations using a paired t-test showing that the Mason-Likar configuration spent 8.2% more time in “green”. There was no difference between electrode configurations in the numbers of false lethal arrhythmia alarms. Abrasive skin preparation did not confer a benefit in signal quality or false lethal arrhythmia alarms. Patients prefer options to carry monitoring equipment. Hairy patients prefer to be shaved to reduce pain at electrode removal.

Implications: There is a difference favoring the Mason-Likar configuration over Lund for mean ECG signal quality, and there is no difference in false lethal arrhythmia alarms. Mason-Likar should remain the choice for continuous in-hospital ECG monitoring. Skin preparation conferred no benefit in signal quality or false lethal arrhythmia detection.

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Chapter 1: INTRODUCTION

Statement of the Problem

Over the past few decades, great advances have been made in continuous electrocardiographic (ECG) monitoring. Improvements in the automated detection of arrhythmias and the addition of ST-segment monitoring for detection of myocardial ischemia have resulted in improved identification of problems, although this technology has added complexity. When the monitor identifies a problem with the patient, an alarm is sounded to alert clinical staff into action. Unfortunately, physiological signals, such as the ECG, are often corrupted with noise. It is believed that noise in the ECG increases false alarms triggering, leading to alarm fatigue in the clinical staff, particularly nurses. The purpose of an alarm system is to communicate information that requires a response or awareness by the operator (Clinical Alarms Task Force, 2007). It has been reported that the rate of false alarms could be as high as 90% (Lawless, 1994; Pahlm & Hammill, 2008; Tsien & Fackler, 1997), contributing to the alarm fatigue phenomenon. The false alarm rate on the ECG is reported to be 42.7% for the most life threatening asystole and extreme bradycardia arrhythmias (Aboukhalil, Nielsen, Saeed, Mark, & Clifford, 2008). Alarm fatigue has been implicated as a contributing factor in the death of hospitalized patients (Kowalzcyk, 2010), and it is reported that the ECG contributes at least one-third of all alarms in an ICU (Chambrin, 2001; Chambrin et al., 1999).

The Mason-Likar electrode configuration has been adopted for stress testing and continuous monitoring because it is perceived to be more noise immune than placing electrodes in standard, distal ECG electrode locations (Mason & Likar, 1966), despite knowledge that torso placed limb electrodes decrease sensitivity for Q-waves of prior myocardial infarction (R. M. Farrell, Syed, & Gutterman, 2008; Kleiner, Nelson, & Boland, 1978; Pahlm & Wagner, 2008a; Zywietz et al., 1990). Controversy exists over the optimal electrode placement for continuous

monitoring, and ECG noise has been cited as a key reason for altering electrode configurations (B. J. Drew & Finlay, 2008; B. J. Drew et al., 1999; Pahlm & Wagner, 2008b; Welinder et al., 2004). An alternate electrode configuration, called the Lund configuration, is more equivalent to the standard 12-lead ECG than the Mason-Likar. It is unknown whether there is a difference between Mason-Likar and Lund electrode configurations if used for continuous monitoring with regard to ECG signal noise, false lethal arrhythmia alarms, and patient comfort (B. Drew, 2011).

A major limitation of prior studies that examined ECG noise was the use of only 10-second ECG recordings rather than continuous monitoring, small sample sizes under 20 subjects, and subjects that were healthy volunteers (Pahlm & Hammill, 2008; Welinder et al., 2004). Compounding this issue is that clinical staff make errors in placement of electrodes, making the issue of standardizing the placement of electrodes important to reduce application errors (B. J. Drew, 2006).

It is obvious that ECG noise can be a source of major negative implications in hospitalized people. First, ECG noise increases the number of false alarms resulting in alarm fatigue, decreased vigilance, impaired patient sleep, and decreased nursing responsiveness to actual alarms (Christensen, 2007; Lawson et al., 2010; Ryherd, Waye, & Ljungkvist, 2008). Second, ECG noise results in misdiagnosis, leading to unnecessary diagnostic tests, inappropriate procedures, and increased length of stay in hospitals. For these reasons, it is imperative to determine whether the Mason-Likar or Lund electrode configurations are different in the amount of ECG noise and false alarms generated.

Purpose of the Study

The purpose of this study is to determine whether Mason-Likar or the Lund electrode configuration represents the ideal configuration for continuous in-hospital ECG monitoring. This comparison will examine the differences for these electrode configurations in signal quality, false lethal arrhythmia alarm frequencies and types, and patient perceptions of each electrode

configuration. Additionally, this study is poised to determine whether an abrasive skin preparation confers any benefit in signal quality or false lethal arrhythmia alarm detection.

Research Aims

Aim 1: In patients in the ICU or PCU, determine whether the Mason-Likar and Lund electrode configurations differ in the amount of myoelectric noise and baseline wander produced in the ECG recordings over a 24-hour monitoring period.

Aim 2: In patients in the ICU or PCU, determine whether the Mason-Likar and Lund electrode configurations differ in the number of total alarms and false life threatening alarms generated by the monitor system's arrhythmia analysis algorithm over a 24-hour monitoring period.

Aim 3: In patients in the ICU or PCU, determine whether an abrasive skin preparation impacts the ECG signal quality and number of false life threatening alarms generated by the monitor system's arrhythmia analysis algorithm over a 24-hour monitoring period.

Aim 4: Describe patient perceptions of each electrode site with regard to comfort, interruption to sleep, and impairment of movement, as well as general perceptions of the experience of being monitored.

Hypotheses

Hypothesis 1: There is no difference in ECG signal quality between the Mason-Likar and Lund electrode configurations.

Hypothesis 2: There is no difference in false lethal arrhythmia alarms between the Mason-Likar and Lund electrode configurations.

Hypothesis 3: There is no difference in signal quality and false lethal arrhythmia alarms with or without abrasive skin preparation in either the Mason-Likar or Lund electrode configurations.

Contributions of this Dissertation Research

There is a major problem with physiologic monitoring alarms in hospitals globally, and the excess numbers of alarms is resulting in patient deaths through alarm fatigue in staff. There are also problems with the use of bedside monitors being used to acquire 12-lead ECG's, but staff are failing to consistently move proximal Mason-Likar positioned limb electrodes onto distal locations proscribed for a standard diagnostic 12-lead ECG, resulting in non-equivalent, nonstandard ECG's being entered into patient records. It has been proposed by ECG experts that the Lund electrode configuration be considered as an alternative to produce bedside monitoring ECG's that are more equivalent to standard 12-lead ECG's than the Mason-Likar acquired ECG's. It is not know whether there is a difference between the Mason-Likar and Lund electrode configurations when used for continuous monitoring in measures of signal quality, false lethal arrhythmia alarms, and patient comfort.

This study is poised to answer the main questions to determine whether the Lund is an acceptable or superior alternative to the Mason-Likar electrode configuration. Additionally, skin preparation has been recommended as a solution to improve signal quality and reduce false arrhythmia alarms, however, this is not known. It is not know what the ideal skin preparation procedure may be, or if any skin preparation technique will prove to be clinically beneficial.

Chapter 2: REVIEW OF LITERATURE AND THEORETICAL FRAMEWORK

Review of Literature: Introduction

Electrocardiology has made great advances since Waller and Einthoven made it possible to sample electrical potentials from the heart at the skin surface of the chest with adequate resolution to determine many aspects of cardiac structure and function (Einthoven, 1924; (Waller, 1887). This paper will serve as a relevant and focused critical review of literature aimed at determining the ideal electrode configuration for continuous in-hospital cardiac monitoring.

There are many aspects to determining the ideal electrode configuration, but it is important first to understand the current state of science. In this paper, three separate domains will be explored that are pertinent to continuous in-hospital cardiac monitoring. The first domain examines similarities and differences between the standard diagnostic 12-lead electrocardiogram (ECG) and continuous ECG monitoring. Included in this area is a discussion comparing two proposed techniques for electrode placement that move limb electrodes proximally. The Mason-Likar electrode configuration is characterized by centrally placed limb leads on the torso, and the Lund configuration with right and left arm electrodes on the outer, lower deltoids and the left leg electrode on the left hip.

The next area to be discussed relates to the skin-electrode interface, since this is where the electrical potentials from the heart are obtained. It is at this connection of technology-to-patient that many questions have been raised regarding the appropriate electrode, conductive materials, adhesives, and skin preparation. Although preliminary research has been done in laboratory settings, measurements from real-world patients have not been accomplished.

The last domain to be addressed pertains to automated algorithms in the monitoring systems, and the alarms created by the algorithms. It has long been proposed that interference or “noise” in the ECG signal causes many false alarms; however, this has never been shown to

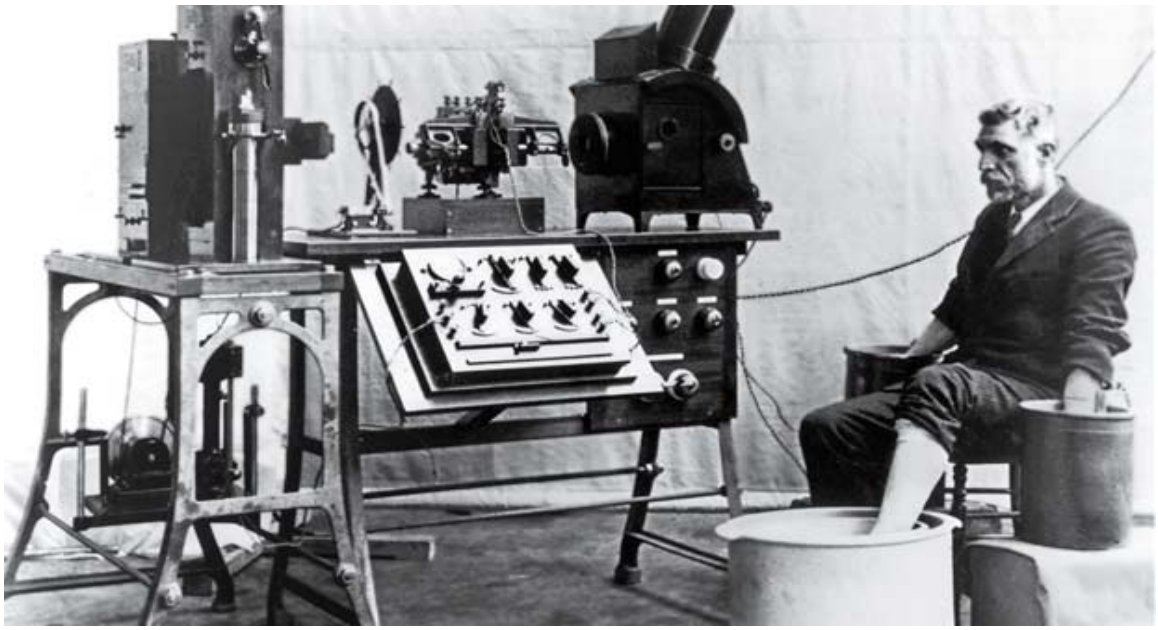
be fact. No researcher has quantified the relationship between signal quality and monitoring alarms. What is known is that clinical alarms are causing alarm fatigue in staff to the detriment of patient care.

By the end of this critical analysis of current and historical literature, the reader should understand the relationships among the three domains of future study: (1) electrode configurations in monitoring, (2) ECG signal quality with particular focus on the skin-electrode interface, and (3) false life threatening monitoring alarms. The next section will then concentrate on relevant theories needed to underpin a conceptual model to more thoroughly understand relationships among electrode configurations, the skin-electrode interface's impact on ECG signal quality, and how this is tied to false monitoring alarms. The measurement and methodology section in Chapter 3 will discuss in great technical detail the techniques used to conduct this research to answer the research questions.

Historical Perspective Primitive Electrocardiogram Acquisition

In 1887, the very first electrocardiogram (ECG) was performed by Waller, who placed the right hand and right foot of a subject into jars of electrolyte solution, connected an electrometer, and observed the movement of mercury with every beat of the heart Waller (1887). Nearly 25 years after Waller, Einthoven published his work and introduced a three lead ECG system. Electrical potential differences were measured between the right upper extremity, left upper extremity and left lower extremity; however, placing subjects' hands and feet into jars of electrolyte solution shown in **Figure-2.1**, but this was obviously highly impractical for clinical use (Einthoven, 1912). In modern practice, conventional acquisition of the ECG still remains to place the adhesive conductive electrodes distally on the arms and legs for diagnostic ECG recordings.

Figure 2.1 Einthoven string galvanometer for ECG circa 1912. Source: Hulton Archives, Getty Images, Downloaded from Wikipedia, Nov. 2013



Development of the Wilson Central Terminal (WCT)

While Einthoven continued his work to better understand the clinical relevance of the ECG, Wilson and his team proposed the concept of the "central terminal" (Wilson, 1934). The Wilson Central terminal is a virtual reference point that is created by averaging the potentials of the left arm (LA), right arm (RA), and left leg (LL). Since the Wilson Central terminal possesses a relatively constant potential throughout the entire cardiac cycle, a waveform that is derived by using the Wilson Central terminal and another electrode will differ mainly because of that particular electrode. It is by this mechanism that modern precordial chest leads are physically unipolar but use the Wilson Central terminal as the second pole in determining a potential difference. A standard diagnostic 12-lead ECG uses a series of six precordial electrodes placed along the anterior chest wall to the left mid axillary line at proscribed locations using ribs, the sternum, and the clavicle as major bony landmarks. The use of six precordial electrodes plus the four limb electrodes means that we acquire a standard diagnostic 12-lead ECG using only 10 total electrodes. There is a fourth limb lead in addition to the LA, RA, and LL that is typically

labeled as RL, which stands for “reference lead”. For simplicity, many technicians explain to novices that RL stands for “right leg” to encourage accurate placement. This grounding electrode is a safety feature with a zero electrical potential that can eliminate extraneous exogenous electricity. In hospital monitoring, wearing all 10 electrodes is presumed to be cumbersome for patients, and frequently the electrodes are in the way of diagnostic tests such as echocardiograms. The additional precordial electrodes can be a nuisance, as well as making it difficult to lie on the left side. For these reasons, it is common practice to use only one precordial lead, typically lead V1 for diagnosing wide QRS complex tachycardias.

Clinical Use of the ECG

In clinical use for more than a century, the standard diagnostic 12-lead electrocardiogram (“standard ECG”) remains one of the most important and the most frequently ordered diagnostic tests in modern medical practice. Many years of population studies have derived normal measurements on the ECG so that it is a meaningful test for diagnosing and monitoring arrhythmias, myocardial ischemia, changes to anatomy and structure of cardiac chambers, and detection of drug-related changes affecting cardiac conduction and ventricular repolarization.

A standard ECG acquires 10 seconds of data, but the utility of a standard ECG for real time ongoing monitoring is impractical for several reasons. Limb electrodes on distal extremities are bothersome for patients in the hospital and contribute enormous amounts of motion artifact to the ECG signal making the waveform nearly uninterpretable. Wires connected to the electrodes are typically stiff due to plastic shielding for durability, and the wires frequently become tangled.

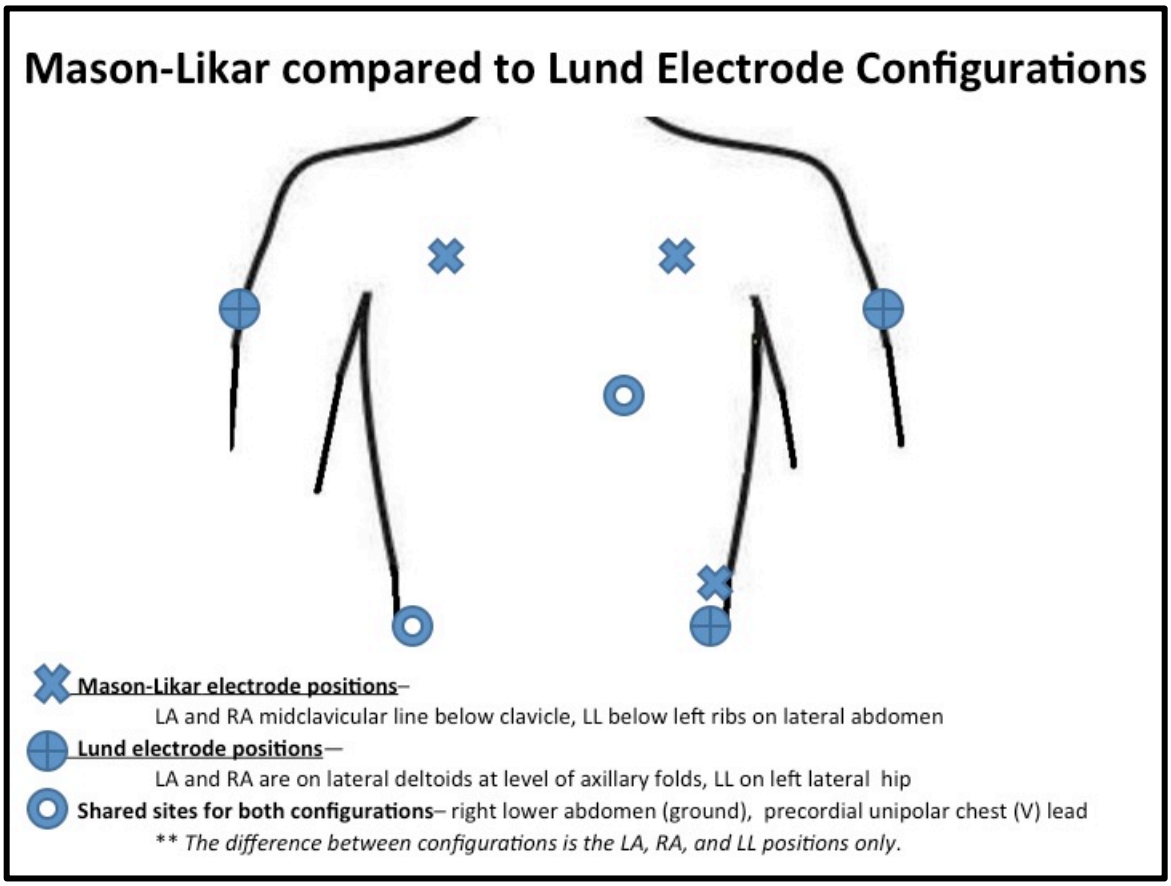
Exercise treadmill stress testing is another frequent use of the ECG for real-time arrhythmia and ischemia detection using all 12 leads for analysis. In 1966, Mason and Likar proposed moving the right and left arm electrodes onto the chest while also moving the left leg

electrode onto the lower abdomen (Mason & Likar, 1966). They hypothesized that this strategy of proximal electrode placement would reduce or even eliminate most of the motion artifact produced by patients running on a treadmill. Although this electrode configuration was originally intended for cardiac exercise treadmill stress testing, the Mason-Likar (ML) electrode configuration was also introduced for continuous in-hospital cardiac monitoring. To this day, patients undergoing cardiac stress testing and continuous in-hospital heart monitoring have their electrodes placed in the Mason-Likar configuration, despite the fact that researchers have identified significant differences between a standard ECG and a non-standard ECG such as the ML configuration.

Alternative electrode placement schemes

It is always preferable to obtain a diagnostic 12-lead ECG with standard electrode positions. This means that the right and left arm electrodes are positioned distal on both arms below the level of the axillary folds, the left and right leg electrodes are positioned at least onto the femur (thigh), and all six precordial leads in the proscribed positions. For stress testing and continuous monitoring applications, it is understandable that clinicians would want to reduce motion interference and artifact in the waveform by moving the electrodes more centrally. Currently, the two alternate centrally positioned systems are the Mason Likar (ML) and the Lund configuration as shown in **Figure-2.2**.

Figure 2.2- Mason-Likar and Lund Electrode Configurations Compared



In the ML configuration, the right and left arm electrodes are proscribed to the infraclavicular fossae on the anterior chest wall and the left leg electrode to the lateral abdomen above the iliac crest but below the lowest rib in the anterior axillary line (Mason & Likar, 1966). Alternatively, the Lund electrode configuration proscribes the right and left arm electrodes on the arms to a point at or below the level of the axillary fold on the lateral deltoids, while leaving the left leg electrode at or below the iliac crest on the left leg or thigh (Edenbrandt, Pahlm, & Sornmo, 1989; Pahlm et al., 1992). In both the Lund and ML electrode configurations, the precordial electrodes remain in the identical locations as the standard ECG.

Ideal Electrode Configuration Evaluation Tool

Prior to adopting a new technique for acquiring ECGs, it is important to ensure that an ECG acquired from an alternative electrode configuration would provide the same data for comparison as the standard ECG. If an ECG were acquired from a non-standard electrode configuration, it would be important for clinicians to be aware that subtle but significant differences may exist in the non-standard ECG. Technology now exists that allows a patient undergoing continuous ECG monitoring to have the six precordial electrodes added to the monitoring cable. Since most patients are being monitored with the limb electrodes in the ML configuration, not the standard distal locations, a non-standard ECG may be obtained and erroneously labeled as standard.

Recent publications have suggested that standardization of electrode placement for continuous patient monitoring would also be important since often times, an ECG may be obtained simply by adding precordial electrodes to existing electrodes with limb electrodes in a monitoring (not standard) electrode configuration. A proposed assessment tool has been promulgated and suggests that several important factors must be evaluated in considering an alternate electrode configuration (B. Drew, 2011). These criteria for evaluating an alternative electrode configuration would include the following criteria:

1. Produces ECG equivalent to the standard diagnostic 12 lead ECG.
2. Provides for patient comfort by not placing electrodes on bony prominences or uncomfortable locations.
3. Produces a non-noisy ECG that does not trigger excessive monitoring alarms.
4. Does not interfere with clinical interventions that must be performed as part of routine care.
5. Must be easy for clinicians to locate anatomical landmarks for consistent and accurate electrode placement.

There are several reasons why use of an alternative electrode configuration may impact clinical practice and patient outcomes. First, It is unclear whether most clinicians know the type of ECG acquisition being reviewed and the implications for a non-standard ECG. Second, many clinicians are unaware of the waveform differences between a standard and non-standard ECG with limb electrodes in more proximal locations. Most non-standard ECG's are not marked as such, and clinicians unknowingly apply the same diagnostic criteria to non-equivalent ECG waveforms (Jowett, Turner, Cole, & Jones, 2005). Lastly, these factors can easily lead to misinterpretation of ECG findings with a risk for misdiagnosis and confusion related to the patient's clinical circumstances. Ensuring methodological integrity in acquiring the ECG for comparison is paramount. Knowing that distal limb electrode placement is often impractical for clinical use, optimizing a more proximal limb electrode location without sacrificing diagnostic quality would have great value in improving diagnostic performance of the continuous ECG, especially when adapted to obtain a diagnostic 12-lead ECG. Systematically, it is important that a convention is developed so that interpretation criteria can be equally applied to all ECGs.

Computer Search Strategies

A preliminary PubMed search began in November 2010, including only citations that were published in English without historical time limitations. There were three domains that were searched for their particular relevance to the research concept: (Sanford P. Bordeau (1982) Volts to Hertz...the Rise of Electricity. Burgess Publishing Company) the Mason-Likar and Lund electrode configuration differences; (Sanford P. Bordeau (1982) Volts to Hertz...the Rise of Electricity. Burgess Publishing Company) skin-electrode interfacing and ECG signal quality; and (3) clinical monitoring alarms. Initial key word search for “Mason-Likar” returned 32 citations, all of which were reviewed but only 9 were considered relevant. Two of these citations were the original published descriptions of the Mason-Likar and Lund electrode configurations (Edenbrandt et al., 1989). Probing the references of these journal articles, an additional 16

references were discovered to be of direct relevance. Additional PubMed and Google Scholar searches for “(bicycle ergometer OR Lund) AND electrode” revealed 32 additional articles with only 5 relevant articles discovered.

The next search began in PubMed and Google Scholar with “skin AND electrode”, returning 4353 articles requiring further refinement to include “signal quality”, which resulted in 47 articles. All 47 were reviewed with only 5 of significant relevance to ECG monitoring. Subsequent reference mining discovered an additional three associated references. Many of these references were discovered but were eliminated due to the experimental electrodes being evaluated, focusing more on commercially available electrodes for ECG monitoring.

The last domain searched related to the clinical monitoring alarms. The search was performed using PubMed, Google Scholar, and Google to capture both professional and mainstream public citations using terms “monitoring alarms AND fatigue”. There were 21 citations that were all reviewed, finding 12 of them directly relevant. Subsequent reference mining was performed on these articles; however, it was discovered that significant cross-referencing occurred. Many related citations were discovered on Google search that were related to monitoring alarm fatigue, but were eliminated if they were only expert opinion or commentary.

Mason-Likar Placement of Electrodes

In the 1960’s, Mason and Likar recognized that early ECG detection of myocardial ischemia and coronary artery disease during stress testing is of major importance. It was understood that if the patient was able to exercise while simultaneously providing 12-lead ECG data, coronary artery disease might be detected by changes on the ECG. Unfortunately, due to motion artifact, ECG waveforms during exercise stress testing were so fraught with artifact that the ECG waveforms were not of adequate quality to be diagnostic. Using P, QRS, and T wave amplitudes as the marker for equivalence, the limb electrodes were moved proximally onto the

chest and lower abdomen (Mason & Likar, 1966). With a sample of 19 subjects, Mason and Likar demonstrated no significant difference in amplitude (as measured in % by height R wave/ depth S wave ratio) in any of the waveforms of the ECG at rest.

The next phase of the study conducted exercise testing using bicycle ergometry on 24 normal subjects and 30 subjects with classic angina pectoris upon exertion. Data showed that placing limb electrodes in a standard position for stress testing detected only 57% of the angina subjects had a positive ECG during stress; however, 80% of those same subjects had a positive ECG during stress while using the ML configuration, presumed to be from better sensitivity associated with improved waveform quality (Mason & Likar, 1966).

Using only ECG waveform amplitude is unsatisfactory for making statements regarding equivalence. A sample size of 19 subjects represents pilot level work by modern clinical research standards. Only 54 total subjects in an experimental implementation group is also a very small sample size considering the large number of variables to be evaluated. ECG equipment, electrodes, and algorithms have changed greatly in the five decades since this study was completed. Examples include improved skin to electrode conductive interfaces, electrode adhesion, improved signal transmission through double braided copper wires, faster signal processing, and electronic data storage. Initially, the ML electrode configuration appeared to be the solution for eliminating myoelectric noise and baseline wander by moving electrodes to a more central location. Although further work was indicated at the time of this research for stress testing, this electrode configuration was rapidly adopted for continuous in-hospital cardiac monitoring without question for more than a decade.

ML Configuration Compared to Standard Diagnostic 12-Lead ECG

Clinicians are unaware of the impact of moving the limb electrodes onto the torso. This section will focus on the critical review of literature related to differences between standard ECG electrode positions and the ML torso positioned electrodes. Compared to the standard ECG,

important differences for ML positioned electrodes have been reported to include the following:

1. Increased QRS voltages in inferior leads (II, III, aVF).
2. Decreased QRS voltages in lateral leads (I, aVL).
3. P, QRS, and T axis shifts toward the vertical (toward either +90 or -90 degrees). The largest axis shifts are on those ECGs where the standard ECG axis is horizontal, about 0°, where the axis shift will be 50° different on average in a rightward direction. For most ECGs, this is a rightward axis shift, but for patients with left axis deviation exceeding -30°, the ML ECG will exaggerate deviation even further to the left (more negative than -30 degrees).
4. A reduction in Q wave amplitude or duration in the inferior leads, leading to a reduced ability to detect inferior myocardial infarction.
5. Generation of abnormal Q waves in inferior leads leading to false positive diagnoses of inferior infarctions.
6. ST voltage deviations from baseline in the lateral precordial leads tend to be mimicked in the inferior leads, particularly in lead III. This effect is dependent on the position of the left leg electrode. Assuming that the left leg electrode is closest to the V5 or V6 electrode, lead III will mimic those leads. If the left leg electrode is closer to V3 or V4, then lead III may mimic these anterior precordial leads.

A decade after Mason and Likar proposed their alternative electrode configuration, several publications emerged supporting the use of the ML configuration, basically endorsing it in modern practice for stress testing (Phibbs & Buckels, 1975; Sheffield & Roitman, 1976). Years passed before any published science suggested that ML ECGs were not equivalent to a standard ECG. Kleiner, et al. (1978) conducted research on 90 subjects. There were 75 consecutive patients and an additional 15 subjects with documented transmural infarctions. All subjects were studied by stress testing with simultaneously recorded ML and standard ECG's.

Of the 75 subjects, 66% had rightward axis shifts of at least 30° in the ML ECG compared to the standard ECG. The mean rightward shift in these 50 subjects was about 45° but frequently exceeded 60°. Importantly, 11 subjects without prior history or ECG evidence of MI developed Q waves and T wave inversions in lead aVL suggesting lateral wall ischemia. Of 15 known transmural infarction subjects, 41% had diagnostic changes of inferior infarction erased by the rightward axis shift on the ML ECG. Subjects with anterior infarction on the standard ECG had the pattern remained unchanged on the ML ECG (Kleiner et al., 1978).

The findings of Kleiner, et al. (Kleiner et al.) were confirmed when 68 more healthy male subjects underwent standard and ML ECG acquisition for comparison. This study demonstrated the change in QRS axis rightward; R-wave amplitude reduction in leads I and aVL, R-wave increases in II, III, and aVF, and changes in the P and T-wave amplitudes similar to the R-wave changes. These researchers also noted significant ST-segment increases inferiorly. Although the ECG changes in this study do not invalidate the interpretation of ST-segment changes during or after exercise, they may obscure or falsely produce evidence of infarction on the baseline exercise tracings (Rautaharju, Prineas, Crow, Seale, & Furberg, 1980). The amplitudes in precordial leads were influenced less than limb leads; however, the R-wave amplitude was reduced in V1 and V2, while the S-wave depth was reduced in V5 and V6 (Rautaharju et al., 1980). These changes were presumed by the authors to be predominantly due to the effect of the left arm electrode

By 1984, another study examined proximal movement of arm electrodes. Gamble et al., (Gamble, McManus, Jensen, & Froelicher) examined 104 subjects with known coronary artery disease, each of whom provided a baseline resting standard ECG while supine then standing. Obtaining a non-standard ECG while supine then standing followed this with electrodes in a modified ML configuration. Placing the right and left arm electrodes 2 cm below the lateral clavicle on the deltoid modified the ML configuration from its original description. Although the findings related to arrhythmia were very similar, the biggest differences among the ECGs

obtained for each subject in this study were the differences between supine and standing rather than between different lead configurations. (Gamble et al., 1984).

Diagnostically, one of the most important differences of an ML ECG is that the inferior surface of the myocardium is not represented in isolation. A high incidence of false negative recordings in patients with true ischemic events occurring only in the inferior wall severely limits the use of the ML ECG for detecting inferior wall ischemia. This shortcoming is compounded by false positive findings in the inferior leads when the true abnormality lies in the anterior or lateral leads (Papouchado, Walker, James, & Clarke, 1987; Toosi & Sochanski, 2008). This misrepresentation of the inferior wall can lead to misdiagnosis and incorrect treatment.

Further comparisons of the ML ECG and the standard ECG continued, and confirmed previous work showing a significant increase in R-wave amplitude in the inferior leads, reduced R-wave amplitude in leads I and aVL, and a rightward shift of the QRS axis. This research introduced the concept that movement of the left leg electrode onto the upper abdomen and lower chest wall on the ML ECG probably represented modified anterior leads (Papouchado et al., 1987). This was concluded because the R-wave amplitudes so closely correlated with those in the anterolateral chest leads. When the left leg electrode is placed at the left lateral lower rib cage, the R-wave amplitude in lead II approximates the R-wave amplitude in lead V5 (Pearson's $r = .7615$, $p < 0.001$). This research also underscored the inability of the stress test to predict the location of coronary artery disease lesions and partly explained a high incidence of false negative tests in patients with inferior wall ischemia (Papouchado et al., 1987; Sevilla et al., 1989).

Acquiring both 12 and 16-lead ECGs in 150 subjects, similar results were found showing rightward axis shifts and voltage changes (R. M. Farrell et al., 2008). This study was unique in that it positioned the electrodes in alternative electrode sites on each subject, such that it was able to be determined which non-standard electrode placement would provide the most accurate approximation of the standard ECG. By testing the potentials at multiple electrode

sites, it was demonstrated that an asymmetric configuration with the right arm electrode on the torso and the left arm electrode on the upper arm may offer some compromise between ECG signal noise and faithfulness to the standard configuration (R. M. Farrell et al., 2008).

In summary, the ML ECG may be useful in some applications but caution must be exercised in others. The ML ECG is useful for global interval measures of the PR, QRS, and QT intervals; rhythm analysis; morphology of the precordial leads; and diagnosis of anterior infarcts whether new or old. The ML ECG should not be used as a direct comparison to the standard ECG in determining any waveform axis, morphology of frontal plane (i.e., limb) leads, and should never be used for diagnosis or monitoring of inferior myocardial infarction. Fundamentally, it is most important that all clinicians understand that direct comparison of standard and ML ECGs is not advisable. It is also important to recognize that although many differences exist between the standard and ML ECG, in continuous in-hospital monitoring situations, a major concern is missed ST-segment changes or false ST-segment alarms that may occur in the inferior leads. The identified ST-segment differences in the lateral precordial leads are of little consequence in continuous monitoring since the AHA guidelines recommend V1 rather than lateral precordial leads (B. J. Drew & Funk, 2006; Gibler et al., 2005).

Lund Electrode Configuration Compared to Mason-Likar and Standard ECG

In 1989, Lund University researchers described a novel electrode configuration that minimized ECG motion artifact while preserving accuracy to the standard ECG for patients undergoing bicycle ergometry stress testing (Edenbrandt et al., 1989). The Lund electrode configuration uses all conventional precordial electrode sites as the standard ECG, but the arm electrodes are positioned onto the lateral side of the left and right deltoids at or below the level of the axillary folds. Although not exactly the same as the standard ECG, the Lund configuration has less pronounced differences than ML ECGs (Edenbrandt et al., 1989). In addition to the arm electrodes, the Lund configuration moves the left leg electrode to the major trochanter of the left

femur to diminish the potential differences between the Lund and the standard ECG (Pahlm et al., 1992).

It is clear that the ML ECG is not equivalent to a standard ECG, and although the Lund electrode configuration appears to be more similar, the impact of adopting the Lund configuration for continuous hospital monitoring is not clear. If we apply the Drew principles proposed for determining the ideal electrode configuration for hospital monitoring, it is imperative that rigorous scientific testing of the Lund electrode configuration be conducted to ensure that this system is equivalent if not superior to other available methods of ECG monitoring.

In practical terms, the Lund electrode configuration needs to be comfortable enough for patients to wear for extended periods of time, diagnostically equivalent to the standard ECG, equal or more noise immune, and have equal or fewer false and/or nuisance monitoring alarms. Several investigators have conducted pilot level research to evaluate Lund ECG's for diagnostic equivalence and noise immunity. For standard ECG equivalence, studies have examined serial standard ECG's compared to Lund ECG's, and found that much like the ML configuration, misplacement of electrodes is often contributory to the erroneous data from the ECG.

A case report of a misplaced left leg electrode too high onto the abdomen/lower torso has been documented to result in incorrect identification of inferior ST changes misdiagnosed as a myocardial infarction (Toosi & Sochanski, 2008). Confusion occurred as these ST changes would appear, disappear, and reappear depending on the placement of electrodes.

Retrospective review of 167 subjects' ECG's demonstrated that the 95% confidence intervals of measurement on the ECGs were more similar when comparing the measurements from the Lund configuration versus the standard recording than when compared to the ML ECG's (Tragardh-Johansson, Welinder, & Pahlm, 2011). This study only reports on the differences between 10-second snapshot ECG's, not on any aspects for continuous monitoring.

Pahlm & Wagner commented on this diagnostic dilemma balancing the issues regarding

patient comfort, standardization between diagnostic and monitoring electrode locations, and marking ECG's accurately so that correct diagnoses can be made effectively. Unfortunately, this publication did not address the huge burden on the medical community to re-educate all providers regarding differences between standard and non-standard ECG's (Pahlm & Wagner, 2008a).

If the Lund electrode configuration is more similar to the standard ECG, then it would seem logical to adopt the Lund electrode configuration for continuous in-hospital cardiac monitoring. Caution should be exercised however, since it is unclear whether differences in noise immunity and patient comfort (tolerability) exist between the currently accepted ML and the proposed Lund configurations.

A small study directly compared ML and Lund ECG's for noise immunity (Welinder, Wagner, Maynard, & Pahlm, 2010). Although the parent study enrolled 80 subjects, the noise immunity sub-study only included 20 subjects, of which nine were healthy volunteers. Each subject contributed two standard ECG's, two Lund ECG's, and two ML ECG's. Subjects were supine and asked to perform leg and arm movements consistent with combing and shaving. Two cardiologists and one ECG monitor technician then performed subjective assessment, providing a 0-5 score for ECG noise. Both the ML and Lund ECG's were compared to the standard ECG's for noise immunity with the arm and leg movements. Comparison of the standard ECG to the Lund ECG shows that the Lund configuration is much more noise immune, while direct comparison of ML with Lund ECGs revealed very similar noise immunity properties. (Welinder et al., 2010). This study is limited by a very small sample size, which was not reflective of the typical status of hospitalized patients. It also used a subjective report of waveform noise from only three raters, and this 5-point Likert scale was further collapsed to a binary "noisy" or "not noisy" rating making parametric analysis impossible. Another limitation for extrapolating this pilot data to continuous in-hospital monitoring is that each subject only contributed a total of 60 seconds of ECG data, which was not continuous or representative of

the normal activities of hospitalized patients.

Current Literature Reflects Inconsistent and Poor Methodology

Analysis of research regarding alternative placement of electrodes deviating from standard ECG distal electrode sites shows multiple gaps in the science. Although many researchers cited previously in this paper described using either a ML or Lund electrode configuration, few of them adhered to the specified electrode locations originally described for each configuration. Diamond, et al (Diamond, Griffith, Greenberg, & Carleton) studied the ML compared to the standard ECG, and described in their methodology misplacing the left leg electrode too high onto the abdomen, which is known to exaggerate axis changes in the ML ECG frontal plane (Diamond et al., 1979). In addition, this study was so underpowered with a sample size of only ten that drawing any conclusions for clinical practice is not feasible.

While comparing the ML and standard ECG similarities, Rautaharju, et al (Rautaharju et al.) also took electrode placement liberties by moving the left leg electrode to a position not described by Mason & Likar, into the anterior axillary line between the costal margin and iliac spine, which is known to change aspects of the ECG (Rautaharju et al., 1980). Gamble et al. (Gamble et al.) moved the upper limb electrodes to the outer clavicle in their attempt to conduct a study comparing the ML and standard ECG, and this unusual electrode placement was later repeated others (Gamble et al., 1984; Takuma et al., 1995). All of this variation in technical methodology leaves the state of the science in a position of lacking technical consistency for comparing results between studies, making it challenging to draw conclusions. Furthermore, since fundamental technical differences exist in ECG acquisition, it seems logical to conclude that a meta-analysis approach to analyzing these data sets would be scientifically unsound.

Studies that have examined the Lund compared to the standard and the ML configurations have also been flawed in design, limited generalizability, and technical flaws with electrode placement. Comparison of the standard, ML, and Lund configurations using serial but

asynchronous 12-lead ECG's were obtained in 75 subjects undergoing stress testing. The analysis only included static, resting ECG's in the supine position in each electrode configuration, not upright with dynamic patient activity. The only analyses conducted used scatter-plots measuring Q-wave amplitude and duration in aVF, with the author's conclusions that the Lund was a better representation of the standard than the ML configuration. This claim is not well substantiated with the data presented in this manuscript (Pahlm, 2008 #23).

The first published report on the use and comparison of the Lund configuration to the Mason-Likar and standard ECG was reported in 1989. This study faithfully used the proscribed locations for each electrode configuration and obtained simultaneous ECG recordings; however this study included only 10 subjects. The ECG data was limited only to resting, supine ECG's that did not represent the dynamic nature of patient movement. Measurement was performed on R-wave amplitudes as was conducted by Mason & Likar (1966), with the addition of QRS axis determination. Although there were statistically significant differences in R-wave amplitudes in the limb leads for both ML and Lund compared to the standard ECG, there were not differences in R-wave amplitudes in the precordial leads. Although both ML and Lund differ from the standard ECG in QRS axis, the Lund differed less than the Mason-Likar configuration (Edenbrandt, 1989 #17). Noise levels in the resting, supine standard, Lund, and ML ECG's was performed by subtracting out the PQRST complexes, performing a RMS (root-mean-square) on the amplitude of the noise, and this was computed for each of the ECG sources. The noise levels were computed on the limb leads (leads I, II, III) and found the noise levels in microvolts to be as follows: Lead I ML 83, Lund 84; Lead II ML 125, Lund 98; and Lead III ML 75, Lund 51. These authors concluded that there was no significant difference in noise levels between the ML and Lund configurations; however, with a sample size of only 10 subjects parametric analysis would be questionable.

In emergency department settings, rapid access to ECG data is paramount, and data was gathered on 30 subjects (10 normal controls and 20 known cardiac disease patients)

examining the anterior acromial placement of arm electrodes, which is neither ML nor Lund recommended locations. The comparative measurement criterion was QRS voltage in limb leads (Takuma, 1995 #26). These researchers found that the acromial placement of arm electrodes produced waveforms “essentially free from myoelectric noise”; however, this was not quantified. The small sample size precluded robust parametric statistical analysis. This article received two subsequently published letters to the editor criticizing the lack of scientific rigor in evaluating the ECG differences.

In summary, it is abundantly clear that nonstandard electrode configurations do not produce the same information as the standard ECG. It is also apparent that the Lund configuration, although more closely resembling the standard ECG, is also not a perfect solution. The ML electrode configuration has been widely accepted for both stress testing and hospital monitoring situations, despite knowledge of nonequivalence. Widespread use of the ML configuration for monitoring coupled with the technology to attach precordial wires to the monitoring cable creates a mechanism for unlabeled non-standard ECG’s to enter patient records. Many clinicians are unaware that differences exist between standard and nonstandard ECGs. What is not known are the ramifications for patient comfort, satisfaction, and wearability for continuous hospital monitoring between the ML and Lund electrode configuration, the impact of the ML and Lund configurations on ECG signal quality, and the relationship that electrode configuration may have on arrhythmia and ST-segment monitoring alarms.

Inaccurate Placement of Electrodes Impacts Interpretation and Signal Quality

It is quite common to see ECG electrodes placed incorrectly. Most clinicians are unfamiliar with significant differences that can occur with relatively small alterations in electrode locations. As described in several studies examining the ML versus Lund configurations, multiple researchers incorrectly described the correct locations for each electrode configuration with drawings illustrating the mistakes (Diamond et al., 1979; Gamble et al., 1984; Kleiner et al.,

1978; Papouchado et al., 1987).

Electrode placement research was conducted on 30 ECG technicians, quantifying accuracy of placement of electrodes for diagnostic 12 lead ECGs (Wenger & Kligfield, 1996). On average, each technician typically performs about 30 electrocardiograms per day (Wenger & Kligfield, 1996). Over half of the time, the V1 and V2 electrodes were placed too high and lateral, while 30-50% of the time leads V4-V6 were placed too inferiorly and leftward. This type of information has been confirmed subsequently in a study examining 120 nurses, physicians, and technicians who were asked to place electrodes on a drawing of a person (Rajaganeshan, Ludlam, Francis, Parasramka, & Sutton, 2008). The results showed cardiologists misplaced electrodes more frequently than nurses, technicians, and non-cardiologist physicians. Additionally, V1 and V2 were commonly placed too high, but in direct conflict with the (Wenger & Kligfield) (Wenger & Kligfield) study, the lateral V4-V6 leads were placed too superiorly. The recommendations jointly promulgated by multiple organizations recommend standardization of electrode placement and underscore the importance of continued training and retraining of staff (Kligfield et al., 2007). Although the aforementioned studies focus on misplaced precordial leads, it is conceivable that the same misplacement phenomenon is happening with the limb electrodes, but this is not well known (Kligfield et al., 2007; Rajaganeshan et al., 2008; Wenger & Kligfield, 1996).

There are numerous ramifications of misplaced ECG electrodes. Poor progression of the R wave height and reversed R wave progression could mimic an anterior myocardial infarction. When the V1-V2 electrodes are positioned too far cephalad, the R-wave amplitude is decreased, leading the clinician to believe that there is a problem with R-wave progression, and quite possibly, an rSR' configuration may develop. The R and S-waves also affect criteria for left ventricular hypertrophy in the precordial leads, so accurate placement is imperative.

Overall, electrode placement accuracy continues to be an issue in both monitoring and standard ECG acquisition. Electrode placement variability is a major contributing factor for the

poor ECG waveform reproducibility seen even in the same subject. Considering that accurate electrode placement is a significant and serious issue, introducing another electrode configuration may add confusion and merits further discussion.

ECG Artifact Causes Misdiagnosis and Inappropriate Treatment

There is a growing body of literature likely underestimating the number of patients that have undergone testing, hospitalization, and invasive procedures unnecessarily due to the presence of artifact that mimics arrhythmia on ECG monitors. Inaccurate electrode placement and limb lead reversal have been implicated in limiting the usability of ECG data, or worse, implying arrhythmia and disease states that truly do not exist (B. J. Drew, 2006).

Muscle tremors can produce high frequency interference in ECG signals, which are particularly difficult for algorithms to detect and filter correctly. One type of ECG manifestation has been seen commonly enough in clinical practice to earn a diagnosis in cardiologic literature. Pseudo-atrial flutter is characterized by a low-amplitude, high frequency baseline that mimics the typical atrial flutter waveform characterized at ~300 beats per minute. One patient with Parkinson's disease was referred for cardiology evaluation for atrial flutter requiring life-long anticoagulation, which was later deemed to be pseudo-atrial flutter (Vanerio, 2007). This represents a good example of one common disease state mimicking another on the ECG.

Atrial arrhythmia misdiagnosis has been documented in patients having been placed on antiarrhythmic drugs and anticoagulation (Vereckei, 2004). Several more cases of pseudo-atrial flutter have been reported in patients with transcutaneous electrical nerve stimulators, which produce bipolar current across sensory nerve pathways to habituate and reduce the sensation of pain. These stimulators are programmable to produced impulses at various intensities and frequencies, and when programmed to produce impulses similar to the rate of atrial flutter, confusion occurs (Hauptman & Raza, 1992; Kimberley, Soni, & Williams, 1987; Weitz, Tunick, McElhinney, Mitchell, & Kronzon, 1997).

Atrial arrhythmias may be uncomfortable, progress to hemodynamic instability, and have significant long-term negative outcomes, but are not as immediately concerning as ventricular arrhythmias, which are typically much more lethal in nature. A case series of 12 subjects who were misdiagnosed with ventricular tachycardia (VT) were systematically reviewed, where 7 arrhythmias were observed on telemetry and the other 5 subjects presented with cardiac symptoms (Knight, Pelosi, Michaud, Strickberger, & Morady, 1999). The diagnosis of VT was confirmed by a board-certified cardiologist in 10 of the 12 cases and 7 subjects were given medications inappropriately, 2 patients received a precordial thump, one had a permanent pacemaker implanted, and another had an implantable cardioverter-defibrillator placed.

In a follow up study of physician behaviors and decision-making, 766 physicians were presented two rhythm strips with artifact resembling VT. The rhythm strip was not recognized as artifact by 94% of internists, 58% of cardiologists, and 38% of cardiac electrophysiologists. Additionally, 88% of the electrophysiologists, 53% of the cardiologists, and 31% of the internists recommended invasive procedures for the VT. This study demonstrates that artifact mimicking VT may result in misdiagnosis and invasive medical procedures (Knight, Pelosi, Michaud, Strickberger, & Morady, 2001).

The Monitor as a Source of Artifact

There are many sources of interference in the ECG signal, and sometimes the monitor itself can produce artifacts to interfere with medical devices instead of vise-versa. A case report of a patient undergoing orthopedic surgery developed wide complex tachycardia at a rate of 140 beat per minute (bpm) in the recovery area (Houtman, Rinia, & Kalkman, 2006). The GE Datex-Ohmeda monitoring system was attached to the patient upon arrival in recovery, and moments later she developed the wide-complex tachycardia. Pacing detection was active on the monitor, and the patient was being paced by her VVI-R (Houtman et al.) pacemaker at 140 bpm. After interrogation of the properly functioning pacemaker, it was discovered that the impedance-

based rate-responsive feature of the monitor was using the same 200-microvolt current and 31 kHz frequency settings that the bedside monitor used for respiratory rate monitoring. This caused mixed bio-impedance signals for the pacemaker to interpret generating inappropriate pacemaking.

Medical Devices as Sources of ECG Artifact

Deep brain stimulators have been known to generate interference making the ECG uninterpretable due to 130 Hz interference (Martin, Camenzind, & Burkhard, 2003). Atrial fibrillation and flutter can be simulated and misinterpreted in orthopedic surgery with pulse-irrigating pumps. This is important since arrhythmias during surgery are typically construed to be demand-ischemic in origin resulting in aborting surgery or unnecessary evaluation immediately postoperatively (Toyoyama, Kariya, & Toyoda, 2000). Other equipment such as infusion pumps, lithotripsy equipment, and intravenous fluid warmers have all been demonstrated to produce ECG artifact that has been mistaken for arrhythmia (Kleinman, Shah, Belusko, & Blakeman, 1990; Paulsen & Pritchard, 1988; Schiller, Heerdt, & Roberts, 1988).

Electrode and Skin Related Issues in ECG Signal Quality

Poor quality ECG waveforms present challenges for both human and automated analysis of rhythm, ischemia, and diagnostic parameters such as interval measurements. Artifact can take several forms, and the impact of each type of interference presents its own technical challenge for algorithmically filtering the noise from the signal, or adjusting the interpretation scheme to accommodate for the noise. In this discussion of signal quality, the terms interference, artifact, and noise will be used interchangeably unless specific clarification is necessary.

There are three types of ECG signal noise that occur due to either physiologic or non-physiologic causes: electrical interference typically caused by electric motors and some electronic devices, usually in the 50-60Hz range, baseline wander which is high amplitude, low-

frequency undulations in the ECG baseline typical of rolling over in bed, stretching the arms, or even the with the normal chest excursion of breathing, and (3) myoelectric interference which is caused by muscle movement underlying electrodes and occurs when moving an extremity or shivering. Myoelectric signal noise is the most common, and regrettably, the most technically difficult to eliminate.

For more than four decades, the skin-electrode interface has been known as a source of interference in obtaining quality signals from the surface ECG. Proper skin preparation, use of non-irritating conductive materials, adhesive electrodes, and correct placement of electrodes are all described as essential for routine use of the ECG for heart rate (Hanish, Neustein, Van Cott, & Sanders, 1971). Multiple researchers have used bench and lab techniques to refine and isolate the source of noise in the signals from electrodes placed on the skin, and multiple findings have been reproduced confirming certain aspects of sources of signal noise. Summarizing the findings of key research-based literature includes the following:

1. Electrode metal to conductive gel/liquid contact is NOT the source of artifact in the signals from skin electrodes. It is the conductive gel/liquid-skin interface that creates most of the artifact.
2. Large variations in resistance and impedance to electrical impulses occur over various aspects of the human body, and there are even large variations in the same subject.
3. Skin characteristics are a poorly understood and under-described component of the skin-electrode interface.
4. Not all skin preparation techniques are equally efficacious at reducing skin resistance between electrodes.
5. Although abrasion techniques for skin preparation seem to be the best method, deploying the abrasive technique for research has been inconsistent due to human factors.

6. Wet-prep electrodes produce higher quality signals that stabilize faster and provide a more stable signal over a 24 hour period than do gel based electrodes.
7. Larger surface area electrodes produce better quality signals.
8. It is unclear which skin preparation technique is truly the best for clinical monitoring use.

Abrasive skin preparation could be a major and immediate method to reduce ECG signal noise, improve signal quality, and make automated algorithms more accurate. To date, all studies have serious deficiencies in sampling, methods, design, and interpretation. None of the studies have enough commonality to justify a meta-analysis. Sample sizes range from unreported to a maximum of 120 subjects, with most studies having only 5-20 subjects. Age and gender may also have a significant impact on skin characteristics, and none of the studies included elderly subjects or people under the age of 25. Racial and ethnic differences were not reported, nor did any study examine the impact of BMI on ECG signal quality.

Technical shortcomings are also present in all of the relevant studies. Skin resistance and impedance data were all derived from skin on forearms and backs, neither of which are clinically relevant sites for electrode placement for ECG monitoring. Not using established electrode sites is particularly limiting with respect to known intra-subject and inter-subject variability across various points on the human body (Tam & Webster, 1977). It is known that larger electrode conductive material surface area produces better signal quality, but this has not been quantified thoroughly in clinically relevant monitoring electrode sites (Tronstad, Johnsen, Grimnes, & Martinsen, 2010), nor is it known whether a maximum electrode size exists. Tam & Webster (Tam & Webster) published that electrode movement, horizontal and vertical, was the source of much of the signal noise, so the impact of various types of electrode disc adhesives may play a role in electrode stability and signal quality (Tam & Webster, 1977).

Several investigators have examined sources of signal noise from surface electrodes

and confirmed that movement between the skin and electrodes produces significant artifact (Fitzgibbon, Berger, Tsitlik, & Halperin, 2002; Odman & Oberg, 1982). There are many reports in the literature discussing novel approaches to designing electrodes built into clothing, various conductive electrolyte liquids and gels, and shielding specific to interference type; however, few of these have gained approval of the Food & Drug Administration (FDA) in the United States, and none of them are in widespread use in the global marketplace (de Talhouet & Webster, 1996; Degen & Jackel, 2008; Gruetzmann, Hansen, & Muller, 2007; Lee, Sim, Kim, Lim, & Park, 2010; Patterson, 1978).

Since several studies demonstrated improved signals with abrasive skin prep, it has been assumed that this is the best method for noise reduction in ECG signals. Inconsistencies in the pressure applied, the grit of the abrasive, the number of swipes, and the participants' skin characteristics have made this aspect difficult to control in clinical research studies (Clochesy, Cifani, & Howe, 1991; Medina, Clochesy, & Omery, 1989). Antibiotic-resistant microbes in hospitals, advanced ages of hospital patients, and risk for skin injury are all reasons for abstaining from abrasive skin preparation. Systematic and controlled use of abrasive skin prep yields good results, but a larger improvement in impedance with longer duration using ultrasound preparation has been found, so perhaps skin abrasion is not the optimal skin preparation method (Farinha, Kellogg, Dickinson, & Davison, 2006; Huigen, Peper, & Grimbergen, 2002).

The literature related to the value of skin-electrode interfacing is still in infancy after more than 30 years of research. This paper has summarized the current state of electrode science, but many variables are still missing to formulate the ideal electrode. As the demographics of society change, as well as the priorities for inclusion of a wider variety of sample characteristics, it is apparent that no study has reported enough detail about multiple factors that potentially influence the skin of subjects and the skin-electrode interface. Race, age, ethnicity, age, BMI, sun-exposure, skin thickness, electrolyte and hydration status, and quantification of body hair

are all absent from the literature. Furthermore, an abrasive skin preparation technique to improve ECG signal quality is concerning to implement in practice standards without short and long-term safety and outcomes data.

Relationship Between Signal Quality and Arrhythmia Interpretation

Interpreting ECG waveforms in the presence of signal noise has always been a challenge for humans and computers alike. Signal averaging provided a novel smoothing technique to allow automated analysis in the presence of signal noise. Although novel for 1990, smoothing filters out subtleties and details in the ECG waveform (Zywietz et al., 1990). The signal averaging approach to automated analysis is obsolete, since computerized signal processing is faster and signal resolution is greater. Also, signal-averaging techniques limit real-time beat-to-beat analysis essential for accurate and timely arrhythmia and ischemia detection.

Analysis of 120,698 ECG's was performed by automated computerized analysis using the GE Healthcare 12SL analysis program (R. M. Y. Farrell, B.J., 2004). All ECGs in the sample had an expert human over-read by a cardiologist to serve as the standard for comparison. ECG signals were analyzed for signal quality using a green-yellow-red scheme, where "green"=clean signal, "red"=high artifact level, and "yellow"= some artifact but interpretable. Although only 5% of the ECGs fell into the red or yellow category, the large sample of ECGs made statistical analysis possible. When the signal quality was interpreted as "green", the discordance between cardiologist interpretation of rhythm and automated interpretation of rhythm differed only in 3.9% of ECGs. When the signal quality was "yellow", discordance in rhythm interpretation rose to 7.4%, and as high as 12.1% when the signal was deemed "red".

This research demonstrated that noise in the signal is a barrier to the accuracy of computerized interpretation of rhythm disturbances. This study used 10-second standard ECG's, but it is the only study to quantify any relationship between signal noise and arrhythmia misinterpretation. The ramifications for continuous monitoring were not explored in this study,

making it difficult to understand whether an association exists between signal noise and arrhythmia detection alarms in bedside monitors. It is important to consider the context of diagnostic 12-lead ECG rhythms and signals being more stable, whereas monitoring situations provide signals that are more dynamic with patients undergoing treatment or ambulatory worsening signal quality. Monitoring algorithms for the ECG are forced to process the data without all 12-leads. It is also important that arrhythmias such as asystole, ventricular tachycardia, and ventricular fibrillation often resemble noise in the signal or an electrode off, and interpretation of these lethal rhythms in particular may be more difficult for algorithms to detect accurately. The Farrell & Young (R. M. Y. Farrell, B.J.) data are also limited in describing any relationship between signal noise and ST-segment changes for ischemia detection (R. M. Y. Farrell, B.J., 2004).

Farrell & Rowlandson (R. M. Farrell & Rowlandson) conducted further work examining the impact of ECG signal noise on automated processing in 90,000 diagnostic 12-lead ECG's evaluated by the automated GE 12SL ECG Analysis Program (version 21). With the exception of the P-wave axis and PR-interval assessment, measurements were not significantly affected by the presence of noise in the signal (R. M. Farrell & Rowlandson, 2006). Automated ST-analysis was included in this study, and the mean ST measurements were preserved in the face of a noisy signal. Criticism of this work is that these were diagnostic 12 lead ECGs with the benefit of providing all of the leads for analysis, not a limited number of precordial leads like continuous monitoring. Other limitations are that only 2 of the ECGs included in the sample were rated as a "red" level of signal noise, which is likely to not be reflective of real-world continuous monitoring; however, this is not known (R. M. Farrell & Rowlandson, 2006).

Under monitoring for ischemia by deactivated ST-segment alarms occurs in about half of US hospitals. The most common reason given for failure to monitor the ST-segment was the perception that artifact will cause excessive numbers of ST-segment alarms (Funk et al., 2010). If the noise in the signal could be removed, interpretation for arrhythmia and ischemia would be

facilitated for both clinicians and computerized algorithms by improving accuracy. Many scientists, especially from industry, focus on improving the algorithms; however, with imperfect signals to process, improvement has been challenging, continuing to be a major contributor to clinical monitoring alarm fatigue.

Physiologic Monitoring Alarms

News media has recently focused public attention on alarm fatigue as a "crisis", and described numerous cases of patient injury and even death associated with staff becoming desensitized to monitoring alarms (Kowalczyk, 2010, 2011a, 2011b, 2013). Government agencies such as the Food and Drug Administration (FDA) and private organizations such as Emergency Care Research Institute (ECRI) have identified improving medical device alarm platforms as a top priority for 2013, after reports that over 543 hospital patients have morbidity and/or mortality associated with monitoring alarms and alarm fatigue (ECRI, 2013). More than 40 years ago when continuous hospital ECG monitoring was introduced and quickly accepted, the goals of monitoring were primarily to assess heart rate and detect lethal arrhythmias. Modern practice standards include differential diagnosis of arrhythmia, detection of myocardial ischemia, and identification of QT-interval changes that could be a harbinger of deterioration (B. J. Drew & Funk, 2006). It is difficult to place a value on the contributions of continuous monitoring to patient care; however, with every intervention should come an ongoing risk-benefit analysis. Recently published data suggests that telemetry-monitored patients had higher immediate survival rates following cardiac arrest (OR=3.67, p=0.02), as well as a higher survival to discharge (OR=7.17, p=0.01) (Cleverley et al., 2013).

As an integral part of clinical monitors, alarms are incorporated via computerized algorithms to alert staff to changes in patient condition. There are alarms for blood pressure, pulse oximetry, ECG, and invasive pressure monitoring devices. Over time, more technological developments have led to more physiologic processes that can be monitored, each with its own

set of parameters, to alert the caregivers should a vital function deviate. Monitors are programmed to have high sensitivity at the expense of poor specificity. In addition to the physiologic monitor's alarms to alert staff, other devices such as intravenous infusion pumps and ventilators contribute to the cacophony of sound present in clinical areas. The overwhelming number of sounds and visual alerts from multiple sources creates a phenomenon of "alarm fatigue" in clinical staff (particularly nurses) as well as patients.

Alarm Fatigue Not Well Understood

The alarm fatigue phenomenon of clinical staff is poorly understood at this time. It is unclear which factors contribute significantly to developing alarm fatigue, since there are individual differences (such as stress outside of work, hearing impairment), environmental differences such as distance from the source of the alarms, and the problem with a high false alarm and nuisance alarm rate. For twenty years, it has been known that the false alarm rate is likely above 90%, and nuisance alarms likely account for 60-99% of the total number of alarms (Aboukhalil et al., 2008; Atzema, Schull, Borgundvaag, Slaughter, & Lee, 2006; Chambrin et al., 1999; Tsien & Fackler, 1997). When the alarms are truly correct, action needs to be taken very infrequently, usually less than 10% of the time (Atzema et al., 2006; Chambrin et al., 1999; Gorges, Markewitz, & Westenskow, 2009; Lawless, 1994; Tsien & Fackler, 1997). Much of the sparse research surrounding the concept of alarm fatigue has focused on descriptive and observational studies. No randomized, double blind, multi-center clinical intervention studies have been conducted to strategize reducing alarm burden in the context of patient safety and outcomes.

Alarm Reductions Strategies

Reducing the number of alarms has been the focus of some studies. The common philosophy has been that technology has caused the alarm fatigue problem, so technology

should be employed to remedy the problem. Widening parameters around heart rate and blood pressure alarms to decrease sensitivity to inconsequential changes has been shown to be effective in reducing alarm burden (Graham & Cvach, 2010; Gross, Dahl, & Nielsen, 2011; Rheineck-Leyssius & Kalkman, 1998; Siebig et al., 2010).

Alarm delays

Other researchers have used strategies to incorporate a delay in sounding an alarm condition until the computerized algorithm has had ample time to verify that a problem truly exists and is of clinical consequence. An example of this delay strategy was implemented with pulse oximetry alarms being delayed 6 seconds with a 50% reduction in the number of alarms (Rheineck-Leyssius & Kalkman, 1998).

Non-invasive blood pressure readings have also been studied by this alarm-delay strategy to reduce alarms, and there was a significant reduction in the number of alarm conditions generated (Biot, Holzapfel et al. 2003). There was no difference in staff response times whether a monitoring delay occurred (Biot, Holzapfel, Becq, Melot, & Baconnier, 2003). Widening threshold alarm parameters intuitively makes sense in reducing the number of alarms, so those findings are of no surprise. Incorporating a delay into non-lethal arrhythmia and vital sign derangements superficially seems harmless; however, computerized algorithms have been given the authority to determine and prioritize whether a parameter change warrants alerting clinical staff into action. There are no studies that have shown these strategies to be safe or unsafe, nor have any studies demonstrated a change in outcome for better or worse.

Redundant measures

Another strategy to reduce alarm burden in clinical areas is the concept of redundant measures. Some vital sign data, such as heart rate, can be derived from the ECG, the pulse oximeter, the arterial line or central line. When multiple sensors are capable of detecting the

same parameter, this multisensory approach has been evaluated as a technique to reduce false alarms (Aboukhalil et al., 2008). Using the arterial line as a second measure of perfusion and arrhythmia, it was identified that 42.7% of ECG arrhythmia alarms are false. Use of the arterial blood pressure for arrhythmia validation showed that 59.7% of the false alarms could be suppressed. Coupling arterial line and ECG data reduced ECG false alarm rate from 42.7% to 17.2% (Aboukhalil et al., 2008). Interestingly, these researchers also found that the true alarm identification rate for ventricular tachycardia was made 9.4% worse using this technique; likely due to the computerized algorithm being unclear whether ventricular tachycardia should or should not have a pulse. This may represent an example where human input is urgently needed for over reading, validation, and action.

Impact of monitor watchers

The purpose of monitoring alarms is to use computerized algorithms to detect changes or crisis and focus clinician attention faster. The question was asked whether a human monitor watcher was even necessary if the monitor alarms were sensitive, but this concept still relies on a human interacting with the monitoring equipment to sort out the specificity problem to determine whether the alarm condition is true or false, whether it is of any clinical consequence, and to prioritize if and when any action needs to be taken. The presence of a dedicated human monitor watcher was studied on 2283 patients over two 9-month time intervals to determine any differences in outcomes (Stukshis, Funk, Johnson, & Parkosewich, 1997). There were no significant differences in mortality, unexpected critical care transfer, or occurrence of 3 of the 5 arrhythmias examined. The monitor watchers were associated with fewer episodes of sustained ventricular tachycardia but more bradyarrhythmias, although no explanation was given.

Summary

In summary, the ECG has evolved from only a concept in the late 1800's into the most

commonly ordered diagnostic test in modern clinical practice. Adaptations of acquiring the analog electrical impulses of the heart have evolved over the century of practical application. The ECG was adapted for stress testing and extrapolated to continuous hospital cardiac monitoring without much science. To determine the ideal electrode configuration for continuous in-hospital ECG monitoring, it is important to consider the three domains examined in this critical analysis of literature. Much is unclear about each of these relevant domains to answer the question of what is the ideal electrode configuration for continuous hospital monitoring in the context of electrode configuration, signal quality with special attention to the skin-electrode interface, and clinical monitoring alarms, Of course, issues of patient comfort and tolerability for electrode configurations and skin preparation are very important.

Theoretical Perspectives

Continuous cardiac monitoring has become ubiquitous in critical care and progressive care environments in modern hospital care. A great amount of information is derived from continuous monitoring, creating a reliance on the monitoring technology to alert staff to dynamic patient status changes. This abundance of data presents risks and benefits for staff that process it, and the patients whose care depends on accuracy. In addition, clinicians must sort and prioritize this monitoring information. This interaction with monitoring equipment can serve as a source of information as well as a source of distraction from other patient care duties.

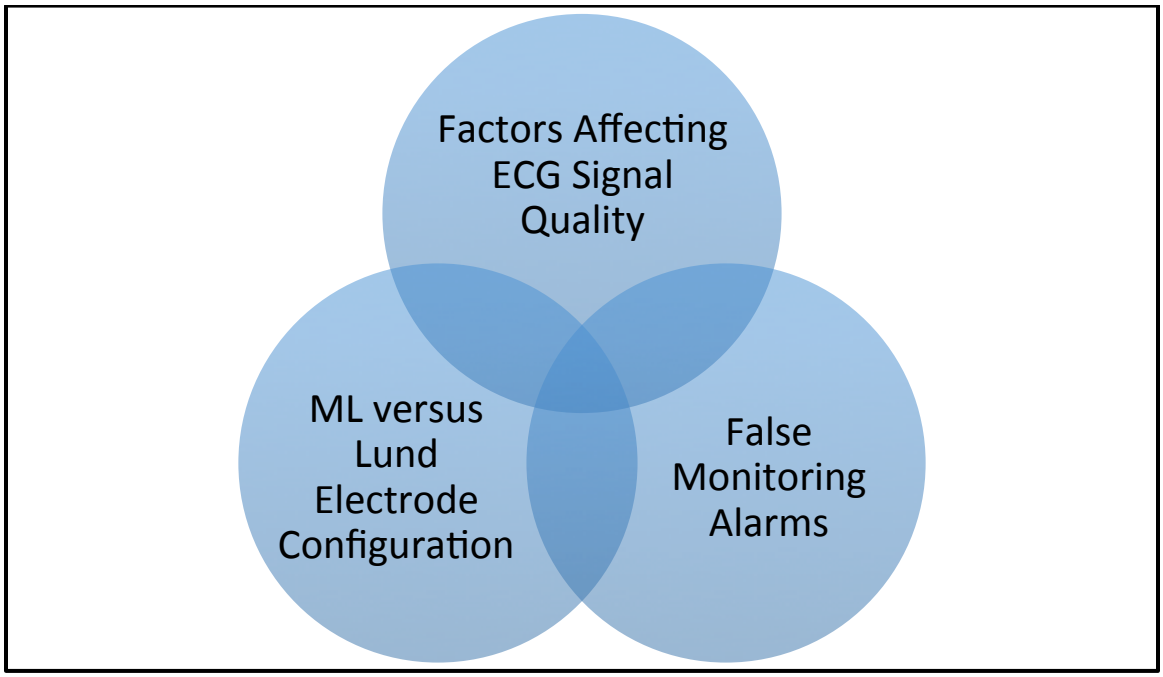
The known limitations of the monitoring technology to reliably produce alarms that are relevant and true is also problematic, manifested as alarm fatigue. The source of many of the automated inaccuracies is the same for human readers. Patients frequently undergo unnecessary testing, evaluation, and even invasive procedures due to inaccurate interpretation of ECG waveforms. As discussed in the review of literature, there are many instances where a noisy ECG signal has led to misdiagnosis and clinical mismanagement. For automated ECG interpretation, a noisy signal is believed to be the source of monitoring alarms and a high false positive alarm rate; however, this is not known.

This paper discusses the biophysics of ECG signal acquisition and processing, electrode configurations, and false monitoring alarms from two theoretical perspectives. First, Ohm's Law will be used as a basis of discussion for ECG signal issues. The second theoretical model is information theory applied to the information gathering, processing, and dissemination process of ECG monitoring. Lastly, both of these theoretical models culminate in the Fidler-Drew Conceptual Model to explain the interplay among ECG signal noise, ECG interpretation, and patient outcomes.

There are three separate and presumably related domains that intersect and interact in determining the best continuous cardiac monitoring practices. Each of the components in this model has its own theoretical underpinnings, which will culminate in an evolved conceptual

model to better understand the interplay of the domains (see Figure 1).

Figure-2.3-- Interplay of Signal Quality, False Alarms, and Electrode Configuration



This paper will begin with a discussion basic cardiac electrophysiology, since electrical potentials of the heart being measured at the skin create the ECG. Next, a description and application of Ohm's law as a theoretical basis for understanding the factors affecting signal quality will be presented. Much of the literature surrounding signal quality relies on understanding impedance and resistance to the flow of micro currents from depolarizing myocytes as the source of much signal noise.

The standard diagnostic 12-lead ECG is obtained by placing the six precordial electrodes in the proscribed precordial locations, the arm electrodes on the arms below the level of the axillary folds, the left leg electrode on the left thigh, and the reference lead anywhere on the body (B. J. Drew & Funk, 2006). To reduce myoelectric noise related to patient movement,

competing electrode configurations have emerged attempting to gain better quality ECG signals. The Mason-Likar (ML) electrode configuration deviates from the standard ECG by having the arm electrodes moved to the infraclavicular fossae, and the left leg electrode moved to the anterior axillary line between the level of the rib and iliac crest (Mason & Likar, 1966). The Lund configuration was proposed those changes include retaining the arm electrodes on the outer lower deltoid below the level of the axillary fold, and the left leg electrode repositioned to the lateral hip over the bone (Edenbrandt et al., 1989)

The interaction of clinicians with monitoring equipment will be described in the context of information theory. Information theory fits well to conceptualize both clinicians' ability to process the deluge of information from physiologic monitors and false monitoring alarms. This theory will be applied to sorting, prioritizing, ignoring, and multi-tasking, while coping with errors in information delivery that are inherent with all communications (Pierce, 1980). This paper will end with a presentation of the original and revised Fidler-Drew Model for conceptualizing the ideal electrode configuration for continuous in-hospital cardiac monitoring.

Cardiac Electrophysiology

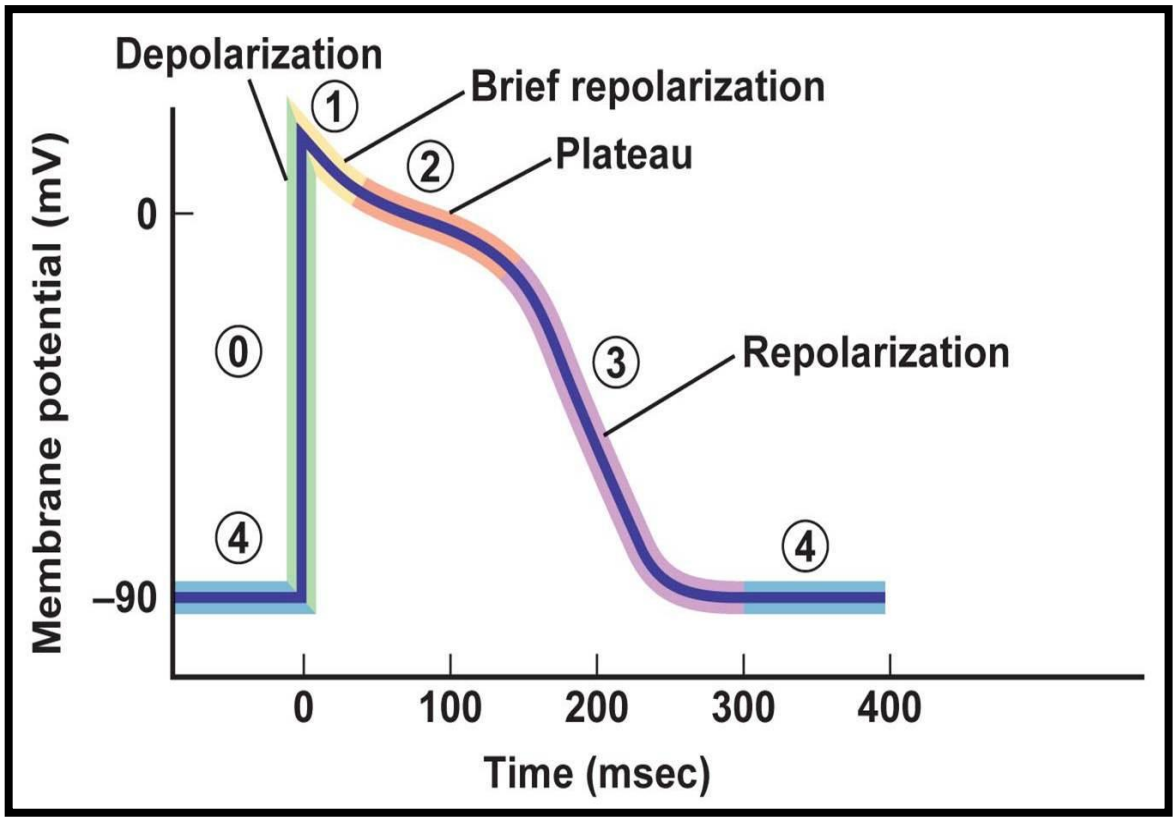
Obtaining the Electrocardiogram

In the most basic sense, the ECG is the measurement of electrical potential differences from the heart measured at the skin surface. If the human heart was perfectly round, and all of the myocytes depolarized from endocardium to epicardium, all of the electrical forces would cancel out any electrical potential difference. There would be no way to record these electrical differences at the skin surface in the form of an ECG waveform. However, the heart is asymmetrical is shape with unequal chamber thicknesses and sizes, and forces do not cancel out. Therefore, differences in electrical potential with respect to time can be recorded on paper as an ECG (Evans, 2006).

Basic Cellular Cardiac Electrophysiology

Activation of the cardiac conduction cells with propagation of action potentials into myocytes is dependent on voltage-gated ion channels in the sarcolemma. These channels manifest very rapid changes in permeability to a variety of ions during the action potential, particularly sodium, potassium and calcium. The shape of the sodium channel is similar to a pore surrounded by a coiled protein-based channel. When activated, as the membrane depolarizes to threshold, this coiled protein channel uncoils, permitting transmission of sodium ions into the intracellular compartment. This changes the cardiac cell resting membrane potential charge from -90mV to +5mV (with a brief overshoot to approximately+20mV) in a fraction of a second, with the movement of electrolytes back to resting state to repeat the cycle (Evans, 2006; Huether, 2004). This process is represented by the cardiac action potential waveform shown in **Figure 2.4**.

Figure 2.4- Cardiac Action Potential (Bell, 2013).



The major phases of the cardiac action potential (see **Figure-2.4** above):

Phase 4: The resting membrane potential is maintained by activation of voltage-sensitive K+ ions, maintaining intracellular negative charge of approximately -90mV.

Phase 0: the phase of rapid depolarization caused by activation of voltage-sensitive sodium (Na+) channels, resulting in a rapid influx of Na+ ions into the cell, creating a relatively positive charge. Typically, intracellular positive charge increases beyond zero up to +20mV. As the sodium ions rush to the intracellular space, an overshoot of electrical charge commonly occurs before the fast sodium channels can close causing the overshoot.

Phase 1: occurs when activation of voltage-sensitive K+ channels produces a net efflux of K+ ions immediately after the phase of rapid activation. This outward K+ current results in transient movement of charge toward partial repolarization of the membrane.

Phase 2: the plateau phase, during which activation of voltage-sensitive calcium (Ca++) channels causes an influx of Ca++ ions, thereby maintaining depolarization charge around +5mV.

Phase 3: the phase of repolarization, caused by activation of voltage-sensitive potassium (K+) channels producing outward movement of K+, Na+, and Ca++ ions. This net efflux of positive charges results in repolarization of the cell back to -90mV.

Loss of Surface Voltage

During invasive cardiac electrophysiology procedures, electrodes placed directly onto endocardium or epicardium produce electrograms of significantly higher voltage than the

surface electrocardiogram. The surface ECG contains very small differences in voltage with respect to time, especially when compared to potentials recorded from an electrode that detects voltage across the cell membrane. Trans-membrane voltages can be as much as 100 mV at the myocardial level, whereas those recorded from the body surface are in the range of 1-2 mV. The ability to deduce what is happening inside the heart using very small voltage changes recorded at the body surface is called the inverse problem of electrocardiography (Evans, 2006). The small voltage potential differences are an important consideration when measuring signal noise, since the electrical potentials of skin and electrode movement frequently exceed the myocardial electrical potentials.

Ohm's Law as Theoretical Construct for ECG Signal Acquisition

A German-trained mathematician and physicist, Georg Ohm did his landmark research on electrical resistance between 1825-1826. Ohm's law is an empirical law generalized from a series of experiments demonstrating that current is approximately proportional to the electric field for most materials (Sanford P. Bordeau (1982) Volts to Hertz...the Rise of Electricity. Burgess Publishing Company). The classic equation to describe the interactive relationship between voltages measured in Volts, current flow measured in Amperes, and resistance to flow measured in Ohms is represented by the equation:

$$\text{Volts (V)} = \text{Current (I)} \times \text{Resistance (R)}$$

Definitions Used in Ohm's Law

Voltage

To thoroughly understand Ohm's law, the definitions of each component of the equation must be accurately defined. An electrical potential is the difference in voltages between two points. For the ECG, voltage (V) is likely the simplest construct to define—this is the electrical

energy potential of the myocardium at various phases during the cardiac electrical cycle. The voltage, as read on the surface ECG, is actually the sum of the potentials of all myocytes electrical activity in the vector toward the surface minus the electrical potentials travelling the opposite direction.

Since more electrical potentials are travelling toward the anterior plane of the human body, the anterior chest wall is the preferred location for electrodes for most routine monitoring and diagnostic ECG purposes (Evans, 2006). In particular circumstances; however, posterior electrodes have utility, such as detection of posterior extension of myocardial ischemia and infarction (Vasaiwala & Schreiber, 2008; Wung & Drew, 2001).

Current

The next variable to define in Ohm's Law is current (I). Electrical current is the flow of electricity through a circuit over a period of time (Hsu, 2005; Sanford P. Bordeau (1982) Volts to Hertz...the Rise of Electricity. Burgess Publishing Company). Electrical current can be described as analogous to the flow of the current of water. With water current, the flow of current occurs downhill, from a source of higher potential energy down to the lower energy location. With electricity, the flow occurs because of an electrical potential difference, from a high or positive electrical potential to a low or negative potential.

Current can be measured several ways, but the typical engineering unit of current measurement is the Ampere, named after the French physicist (Andre Marie Ampere) who first measured electrical flow. Technically defined, the ampere (also known as an "amp"), is the amount of electric charge passing a point in an electric circuit per unit time, with 6.241×10^{18} electrons (one Coulomb) per second constituting one Ampere (Hsu, 2005).

As an example of current, another clinically useful unit of measure is the Joule (J), which is commonly used to measure defibrillation energy transfer. One Joule equals the energy of electric current of 1 Ampere passing through a resistance of 1 Ohm for 1 second.

Resistance

The last component of the Ohm's law equation is resistance (R) measured in Ohms. Defined by the International System of Units (SI), one-Ohm unit is the resistance between two conductor points when a constant electrical potential difference of 1.0 Volts is applied to these points with a current of 1.0 Ampere (Hsu, 2005; Sanford P. Bordeau (1982) Volts to Hertz...the Rise of Electricity. Burgess Publishing Company). Resistance is the reason why myocardial voltage potentials are -90 mV to +20 mV, but at the skin surface, only -10 mV to +10 mV are typically seen on the ECG.

The terms resistance and impedance are frequently used in technical literature describing the skin-electrode interface, but many clinicians poorly understand the difference. Resistance is the opposition to the flow of direct current (DC) electricity, while impedance is the opposition to flow of alternating current (AC) electricity. Both resistance and impedance are measured in Ohms. Alternating current is electricity that alternates polar directionality with alternating positive and negative values. It is typically the form of electricity from a wall receptacle, and in the United States, AC electrical current alternates 60 times per second, or 60 Hertz (Hz) (Hsu, 2005). All electrical potentials generated by the human body are DC electricity; hence, all of the electrical energy lost between the myocardium and the surface ECG electrodes is due to resistance, not impedance.

In the case of ECG monitoring in humans, the amount of resistance between the heart and the skin has multifactorial variation and is not very predictable. Resistance can be increased by systemic electrolyte abnormalities, alterations in hydration status, skin thickness, chest dimensions, adipose tissue, and body temperature. It is also important to know that resistances in an electrical circuit are additive—meaning that the resistance of pericardium is added to that of the costal muscles, rib bones, subcutaneous fat, and skin prior to the electrode Ag-AgCl conductant. An electrical conductant is a medium through which the flow of electricity is established, with the antonym being an insulator, which resists or prevents electrical current

flow (Hsu, 2005).

Poor hydration and dry skin increases resistance, since dry skin has fewer conductive electrolytes present. Dead skin cells are dry, free of electrolytes, and offer great resistance to conducting electricity. Oils and fats are also typically poor conductors of electrical current. These factors support the theoretical connection between skin preparation to remove dead skin cells and oils, while hydrating the skin to reduce resistance (Hanish et al., 1971).

It is important to consider skin resistances when attempting to identify the ideal electrode configuration for continuous ECG monitoring. In addition to the condition of the skin, body hydration status, and skin preparation, various areas of the body have differences in resistances (Huigen et al., 2002). The Lund configuration places the arm electrodes on the outer upper arms and the left leg electrode on the lateral hip (Edenbrandt et al., 1989). The Mason-Likar configuration more centrally places the arm electrodes on the infraclavicular fossae and the leg electrode on the lateral abdomen below the ribs but above the iliac bone (Mason & Likar, 1966). Increasing distance of the electrodes from the heart is known to increase resistance; however, the upper outer arms have a lower skin resistance than the chest (Huigen et al., 2002). It is not known whether there is a difference in signal quality between the Mason-Likar and Lund electrode configurations regarding resistance, when both the distance from the heart and skin resistance variation in electrode sites are factors.

By definition, an electrode that loses contact increases resistance to infinity since there is no longer a flow of current, and conversely, two electrodes connected to each other should exhibit zero or near-zero resistance. Several additional factors that can change resistance to obtaining high-quality ECG signals at the skin surface are distance and electrode lifting. Distance from the source of electrical potentials adds some resistance even with highly conductive wire; so adding distance from the heart to the electrode on a human will add resistance as well (Hsu, 2005). Clinically, a barrel-shaped chest increases anterior-posterior diameter making the skin electrodes further from the heart. The increased distance combined

with increased lung volumes placing electrically insulating air between the electrodes and the heart, makes the ECG waveform amplitudes smaller for barrel-chested patients (Pipberger et al., 1967).

More resistance can occur at the skin-electrode interface where poor adhesion of the skin electrodes can cause movement, particularly vertical lifting from the skin surface to rapidly increase resistance (Kappenman & Luck, 2010). Most research on the quality of electrodes has focused on conductive materials and skin, but the adhesive backing of electrodes is just as important for obtaining high-quality ECG signals. If the electrodes do not adhere well to the skin, both vertical and horizontal electrode movement occurs to degrade signal quality.

The Use of Impedance in Physiologic Monitoring

Impedance measures using AC are clinically used in current ECG monitoring. Through the same electrodes by which the surface DC current from the myocardium is received and then translated into an ECG waveform, small amounts of AC may be transmitted between electrodes to measure impedance without discomfort or harm to the subject. Since intra-thoracic changes in impedance are mathematically predictable, this technology has gained popularity for respiratory rate monitoring and chest compression depth and rate measurement, and may have a role in determining physiologic fluid balances (Donnelly et al., 2013; Gong, Chen, & Li, 2013; Wynne et al., 2006).

Summary of Electrical Physics in ECG Monitoring

Understanding Ohm's law is fundamental to understanding the acquisition of ECG electrical potentials at the skin surface. This law is implicit in quantifying signal quality, noise, and challenges in accurate arrhythmia and ischemia detection. Using Ohm's law as a theoretical concept in ECG monitoring focuses the discussion on the monitoring equipment, the skin-electrode interface, and the electrical conduction system of the heart.

This theory can be applied when considering various skin electrode placement schemes.

It appears that moving electrodes proximally from the Lund to Mason-Likar positions may decrease resistance by both shorter distance and thinner skin thinner on the chest than the arms. However, it is not clear that this change in resistance is actually present or clinically significant. Even if the resistance is less and ECG signal quality is improved, the faithfulness to the standard ECG used for clinical diagnosis is not retained in the Mason-Likar position

Information Theory Applied to Clinical Monitoring Alarms

Humans are constantly presented with information arriving to our nervous system by our senses of sight, smell, sound, touch, and taste. The individual either seeks information or it is organized and presented to them by another source. The other source could be such as another person, book, television, or the Internet. It is often difficult to distinguish information transfer from communication, since communication represents a significant amount of bidirectional information exchange. Information that is never communicated is of little, if any, value. There are multiple phases to information processing in the context of normal daily living, and even more complexity in the healthcare environment as clinicians assimilate information from multiple sources about their patients. In the discussion about clinical physiologic monitoring technology, there are multiple phases in the flow of information between patients, monitoring device, clinicians, support services, and eventually back to the patient.

Information Theory Development

Previously known as communication theory, Harry Nyquist proposed information theory to represent the evolution of communication theory into the mathematically based models incorporating technical aspects of transfer of electricity and electronic signals used in telegraph and telephone transmission (Nyquist, 1924) Nyquist made several major contributions to information and communication theory that are relevant to the topics of this paper. Nyquist described communication as “information”, and suggested that two factors, signal waveform shape and choice of code for transmission, affect the maximum speed of transmission.

Nyquist refined his work theorizing that transmission speed was based on the Sampling Theorem (Nyquist, 1928). Sampling Theorem states that any continuous analog signal sampled at regular intervals over time must be sampled at more than twice the frequency of its highest-frequency component to be converted into an adequate digital form. Thus, the Nyquist Frequency of an analog signal is the highest frequency that can be accurately sampled. However, there is an assumption that there is no data loss upon digital signal reconstruction. Another important concept promulgated by Nyquist states that the signal level must be sufficiently higher than the noise level to maintain acceptable transmission quality (Nyquist, 1928). Later work by Claude Shannon demonstrated that perfect signal transmission is not possible, and errors in transmission are an expectation (Shannon, 1948).

The work of Nyquist and Shannon is directly applicable to ECG acquisition and signal processing. The analog electrical impulse waveforms from the surface ECG used in continuous monitoring are currently sampled somewhere between 125-500 Hz (samples per second), meaning that the ECG waveform produced would be accurate down to half of that sampling frequency. Applying the principles of the Sampling Theorem directly to ECG monitoring raises interesting questions. High-resolution sampling rates of 1000 Hz are now possible due to improved computer technology. Applying the Sampling Theorem to 1000 Hz sampling means that accurate waveform display of ECG impulses of up to 500 Hz may be detected. It is unclear what his increase in resolution of surface ECG potentials would mean clinically. Despite the technological advancement, it is unknown whether signals that are greater than 500 Hz are meaningful on the ECG or have any clinical impact.

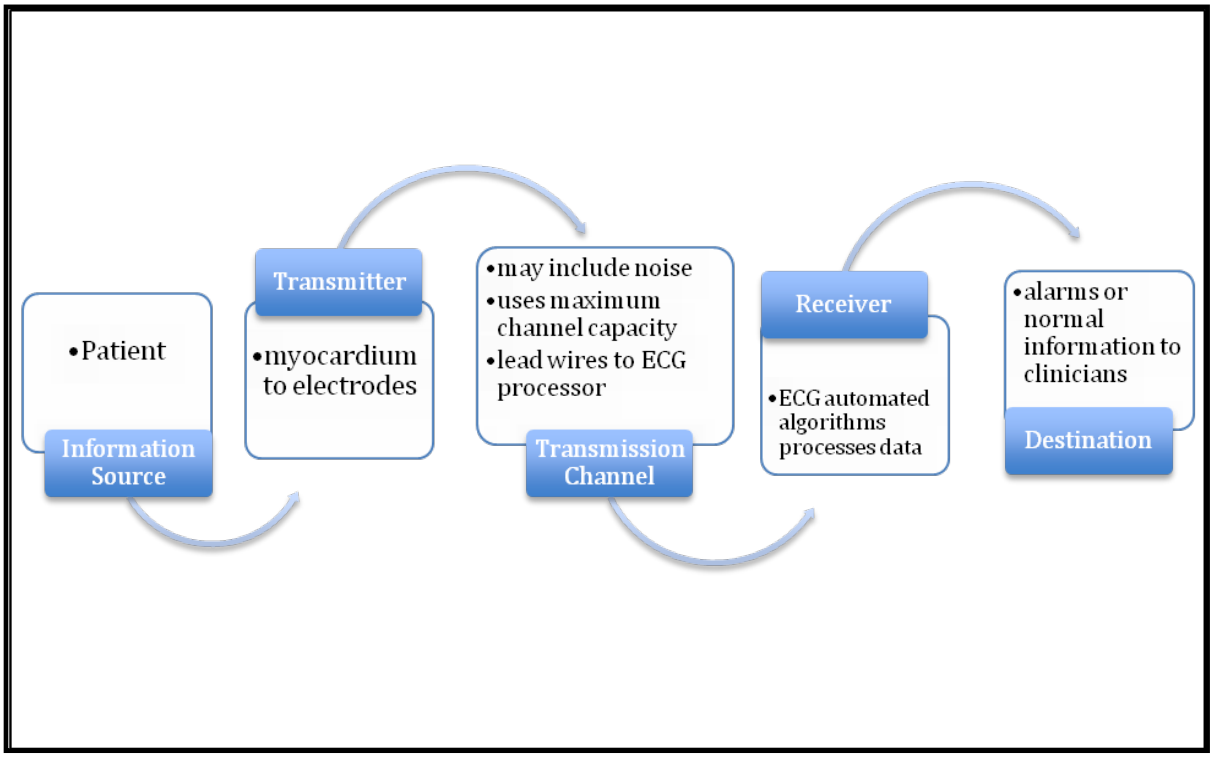
Another concept that reduces the usefulness of Nyquist's hypotheses is imperfect or missing data. Patient movement, poor skin-electrode interfaces and equipment failure such as wire disconnections or dead telemetry batteries, frequently corrupts ECG data. In a controlled laboratory environment, it might be possible to eliminate most sources of artifact; however, in real clinical practice, it is not possible to make the person immobile and eliminate clinical

manipulation. One aspect of this early work by Nyquist, that the signal level that is being measured must be greater than the noise level in the channel of transmission, still continues to be true.

Shannon-Weaver Sender-Receiver Model of Information Processing

The Sender-Receiver model first described by Shannon and Weaver in their work around information theory has typically described communication in global terms (Shannon, 1948). The components of this model depict a systematic process of the sender creating a message that is transmitted with a certain expectation of noise in the signal. On the receiver end, the message arrives and is interpreted by the receiver. The original description of this model portrayed the process as a full loop, where feedback from the destination was given back to the information source.

Figure-2.5--Shannon-Weaver Model to Applied to ECG Signals, adapted from (Shannon, 1948)



Originally, this model was used as a mathematical model of communication, and included tools of probability theory. At that time and even today, the expectation of perfect transmission of information without signal noise, degradation of signal, or errors in encoding information is unrealistic. Shannon believed that all communication is destined to contain some errors and omissions in the original message itself, in the transmitting medium, or the interpretation by the receiver. This built-in expectation of error led Shannon to develop the concept of information entropy as a measure of uncertainty in a message (Shannon, 1948). The concept of entropy relates directly to quantifying the probability of error in the message to be transmitted. Mathematically, the number of data points contained in a message is directly correlated to the probability of error. In ECG signal processing, entropy is introduced by patient movement, clinician manipulation of the patient, and faulty skin-electrode interfacing.

Shannon also proposed the concept of signal transmission capacity. The Shannon Theorem proposes that no matter what is done to the form of the data, the transfer rate cannot exceed the capacity of the channel (Shannon, 1948). The capacity of the transmitting channel is limited by many of its physical properties, such as conductivity and shielding from interference. This remains a factor as manufacturers struggle with the balance between material and production costs, as well as conductivity and signal processing speed. Several aspects that Shannon neglected to address are the receiver capacity, as well as the ability of the destination (clinician) to prioritize and validate the messages from the receiver. Despite increased computing power in the 65 years since the Shannon Theorem was developed, the computer processor receiving ECG information has many difficulties with the accurate interpretation of ECG waveforms for arrhythmia recognition and ST-segment change detection.

Entropy and Signal-to-Noise Ratio

According to the Shannon-Weaver model, if ECG information obtained is corrupt such as with poor electrode adhesion or conduction, the data from the sender already has errors. It is at

this very first phase of the process that an opportunity exists to reduce the error in the information, while simultaneously improving the amount of good data. The best approach to improve the acquired raw data has not yet been determined. In the case of cardiac monitoring this might include choosing better quality electrodes, ascertaining the role of skin preparation, or optimizing the electrode configuration to minimize noise in the signal. To use Shannon's term, entropy of the raw data can be measured in terms of a signal-to-noise ratio. There may be several approaches to improving this ratio, such as increasing the sampling rate, decreasing the error rate, or both.

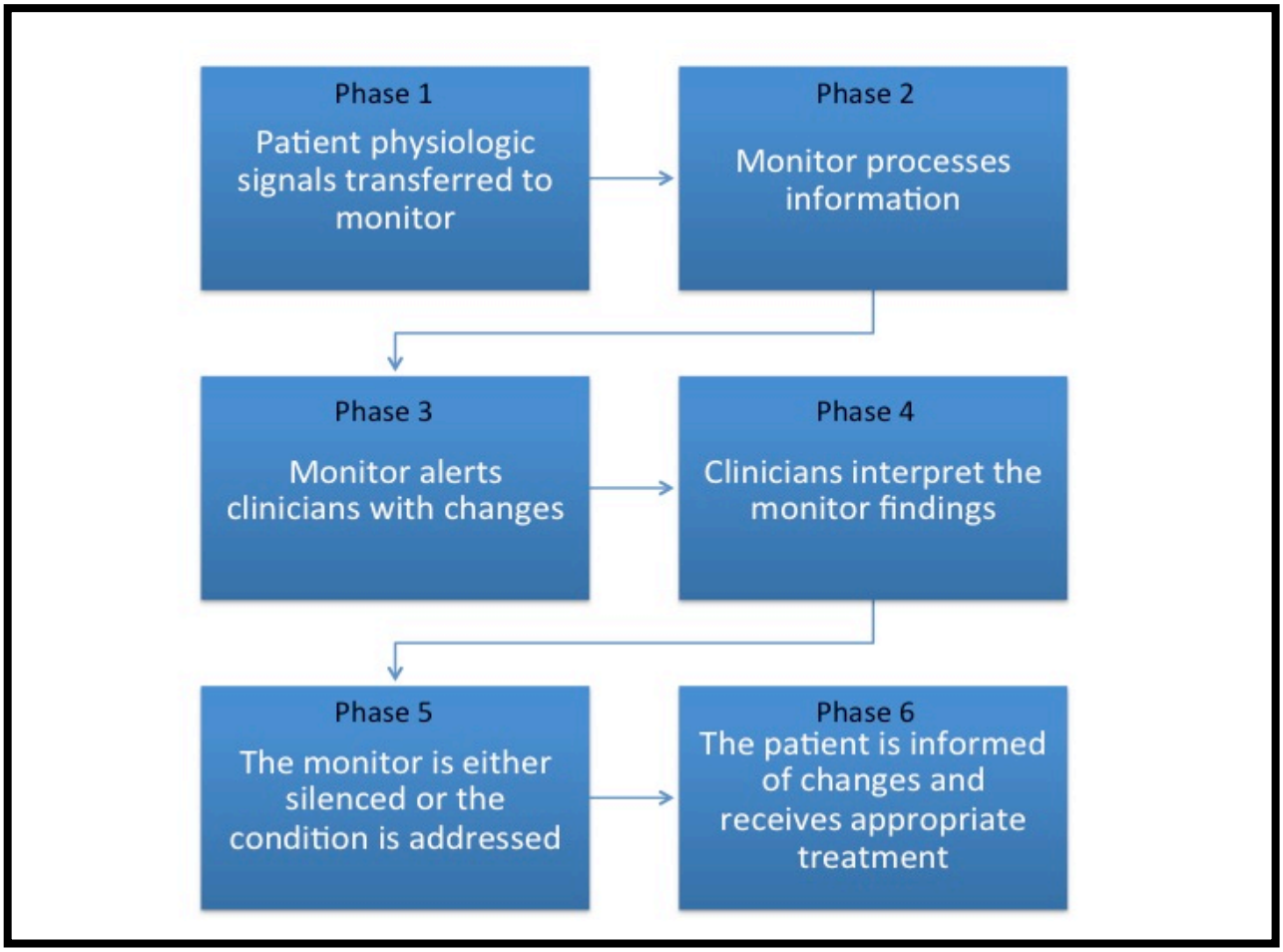
In a simplified example, if the ECG monitor collects 100 bits of data at a sampling rate of 125 Hz, and we expect 5 of those data bits to be missing or erroneous, the signal to noise ratio is 95%. If we discover a method to decrease the error rate to 3 bits at that sampling rate, the signal to noise ratio is now improved to 97%. Some scientists have proposed increasing the sampling rate to improve data granularity, and technology exists to sample at 1000 Hz. According to information theory, information entropy (the chance of error) increases due to the 8-fold increase in data collected over the same time interval. In terms of raw data, the high sampling rate now transfers 800 bits of information to the automated algorithm instead of 100 bits. Even at a 5% error rate, there are still 760 bits of information for interpretation despite 40 erroneous or missing bits. Even though signal to noise ratio in the high-sampling rate strategy is worse than the improved data acquisition technique; there are more bits of data to analyze in the high sampling rate signals. What is not clear in theory or in practice is whether bad bits of data are better, the same, or more detrimental to signal processing than missing data bits.

Information Transmission Error

It is obvious that there are multiple opportunities for information error. Information theory serves as a relevant platform for the representation of the interchange between patients, physiologic monitors, and clinicians, hopefully completing the information and communication

loop to the end receivers—the patients and clinicians. By applying the theoretical constructs of information theory to physiological monitoring, the flow of information can be expressed in a dynamic process of phases (see **Figure-2.6**).

Figure-2.6--Phases of Information Transfer in Continuous ECG Monitoring



Phase 1 of **Figure-2.6** represents signal acquisition and transfer to computer processors. Much attention has been given to various elements of high-quality signal acquisition, including skin-electrode interface, electrode configurations such as the ML and Lund, and methods to reduce noise in the signal prior to arriving at the computer processor.

Interventions such as skin preparation and electrode wire shielding against outside electrical interference also fall into this phase of information transfer for ECG monitoring.

Moving forward, phases 2 and 3 relate to the mechanisms used by the physiologic monitors to process the signals into a waveform or number that is meaningful and useful to the clinical staff. Most of the time, the patient parameters are within normal limits and the monitor remains silent; however, whenever changes occur that either exceed a set parameter or meet other abnormal criteria, the monitor produces audio and visual alerts to the staff. The clinical staff must see and/or hear the alert in order to process the information in the phase 4. It is at this point that opportunities exist for human-technology information lapses to occur. The first lapse occurs with the sensitivity and specificity of the monitoring equipment. It is presumed that bedside physiologic monitors approach 100% sensitivity at the expense of poor specificity. The poor specificity has been documented numerous times (Aboukhalil et al., 2008; Tsien & Fackler, 1997). Corruption of physiologic signals with noise is proposed to be one mechanism for reduced sensitivity and specificity; however, this has not been proven. More often, the monitors create an alert when a true change in status has not occurred, causing a false alarm. The second major lapse in information transmission is alarm fatigue-- not all transmitted information is received, nor is all received information acknowledged.

Phases 5 and 6 in the flow of information represent the human-technology interchange and verification of information, completing the feedback loop from receiver to sender. The constant barrage of information produced by bedside monitors can be overwhelming to even experienced staff; however, the addition of a high false alarm rate creates another layer of complexity in the human psyche when presented with monitoring alarm data (Bliss & Chancey, 2013). The combination of frequent delivery of information, high false alarm rates, combined with the corruption of that data making interpretation a time-consuming challenge (Bliss & Dunn, 2000). It is not currently well described what individual factors place a clinician at risk for alarm fatigue, nor is it described what monitoring alarm load can typically be managed by the human

sensorium before information neglect begins (Bliss & Acton, 2003; Bliss, Fallon, & Nica, 2007; Bliss, Liebman, & Chancey, 2012).

Application of Theory to Understanding Continuous Cardiac Monitoring

Ohm's law sets the stage for understanding the complexities and challenges associated with acquisition of a good quality ECG signal that has minimal noise. Then, in turn, information theory can be used to clarify the flow of information from patient to monitor, monitor to clinician, clinician to monitor, and clinician to patient. Since this interplay has not previously been examined from these theoretical perspectives, a new conceptual model has been developed by Fidler & Drew to conceptualize the relationships surrounding ECG signal quality, monitoring alarms, clinician behaviors, and patient outcomes. Both the original version of the conceptual model and the revised model will be presented in this section.

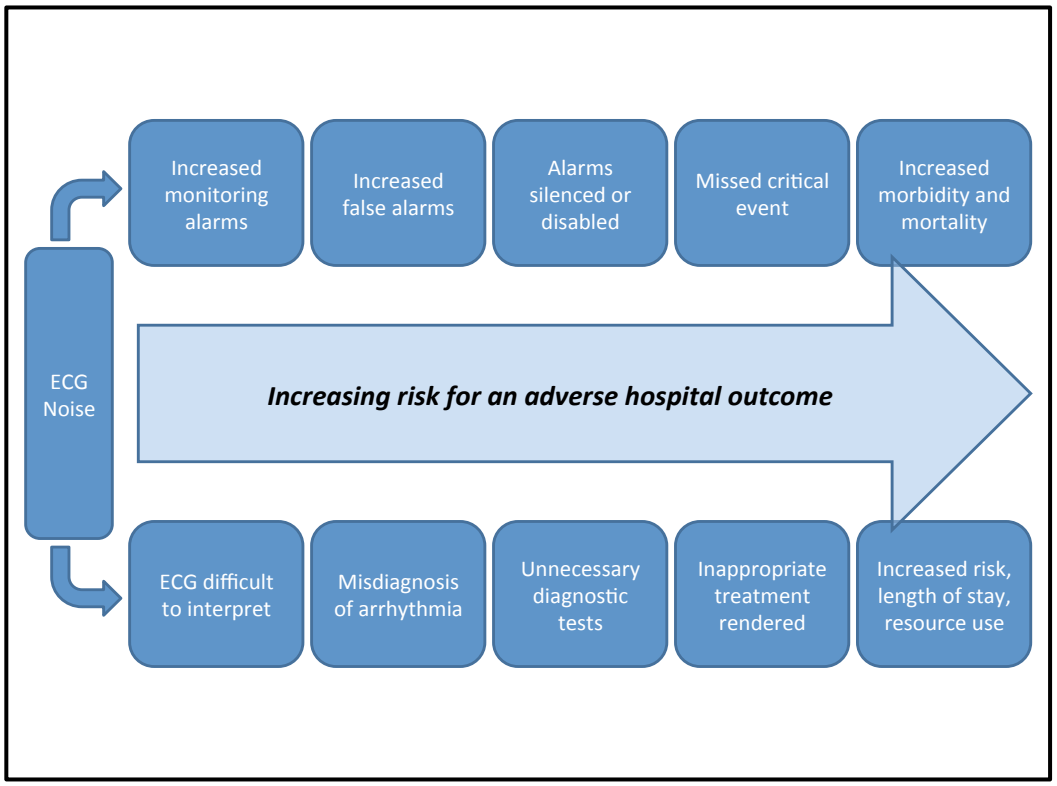
Using suboptimal ECG signal quality as the starting point, this model describes dual mechanisms leading to worsening patient outcomes in monitored hospitalized patients. Despite published practice standards, the acquisition of ECG signals is variable at best (B. J. Drew & Funk, 2006). It is known that clinicians have difficulty placing electrodes in correct locations, but it is not known whether there is a difference between the Mason-Likar and Lund configurations regarding noise in continuous monitoring situations (Pahlm & Wagner, 2008a; Welinder et al., 2010; Wenger & Kligfield, 1996). In the standard diagnostic 12-lead ECG, the presence of noise in the ECG signal impairs automated interpretation mildly, not commensurate with the amount of arrhythmia and ST-segment alarms found in continuous monitoring (R. M. Y. Farrell, B.J., 2004). The Fidler-Drew model depicts the proposed cascade of events in both alarm fatigue development and misdiagnosis increasing the risk for an adverse hospital outcome.

Conceptual Model for Understanding Signal Noise and False Monitoring Alarms

Original Fidler-Drew Model of the Impact of ECG Noise

As depicted in the original Fidler-Drew Model shown in **Figure-2.7**, it is believed that noise in the ECG signal worsens data quality for both automated machine and human analysis. The fundamental cause of both arms of this model stems from ECG signal noise. This model suggests that a noisy ECG signal results in two distinct pathways leading toward worsening hospitalized patient outcomes. On the upper arm of the model, ECG noise is implicated in the process of developing alarm fatigue in clinicians, particularly nursing staff. The lower arm of this model depicts the direct ramifications of ECG signal noise on human ECG readers' ability to correctly diagnose arrhythmia and ischemia information, and subsequent inappropriate treatment of patients.

Figure-2.7-Original Fidler-Drew Model of the Impact of ECG Signal Noise



Beginning with the upper arm of this model, alarm fatigue has been previously described as a process of desensitization to clinical monitoring alarms. Although many medical devices have alarm features built into their controls, this paper focuses on the physiologic monitoring alarms only, particularly the ECG alarms. Hundreds of patients have been reported as being harmed by a lack of response by clinical staff to monitoring alarms (Alarms Task Force, 2007; Cvach, 2009; National Clinical Alarms Survey, 2011). In this model, it is proposed that a noisy ECG signal results in both increased absolute numbers of alarms as well as an increase percentage of false alarms, although this has not been proven. Also not known is whether alarm fatigue is a function of the number of alarms, false negative rate, or a combination of both (Bliss & Dunn, 2000). It is also poorly understood what individual physical and psychological factors in clinicians (examples: amount of sleep prior to working, auditory and/or visual sensory impairment, and psychological mood) contribute to the development of alarm fatigue (Bliss & Acton, 2003; Bliss & Chancey, 2013).

As a patient deteriorates, multiple physiologic parameters may become abnormal as their condition declines. As these physiologic parameters are violated, the monitors at trigger alarms. Some patients that are not deteriorating may also exhibit a high alarm burden due to ECG signal noise, such as patients with combative altered mental status, shivering, and poor skin integrity. Typical responses by clinicians responding to alarms are to first silence the alarm and then determine what caused the alarm state (Bliss & Dunn, 2000). If no cause is found, staff may even disable the alarms not recognizing the patient's deteriorating condition. This can be very dangerous for the patient to have their deterioration occur without the monitor being able to alert nurses that problems exist. The missed critical events have been featured in numerous media venues (Kowalczyk, 2010). Missed critical events jeopardize patient lives every time the alarms are turned off, but it is also difficult for clinicians to differentiate between patients who are stable with high monitoring alarms rates and the truly deteriorating patient.

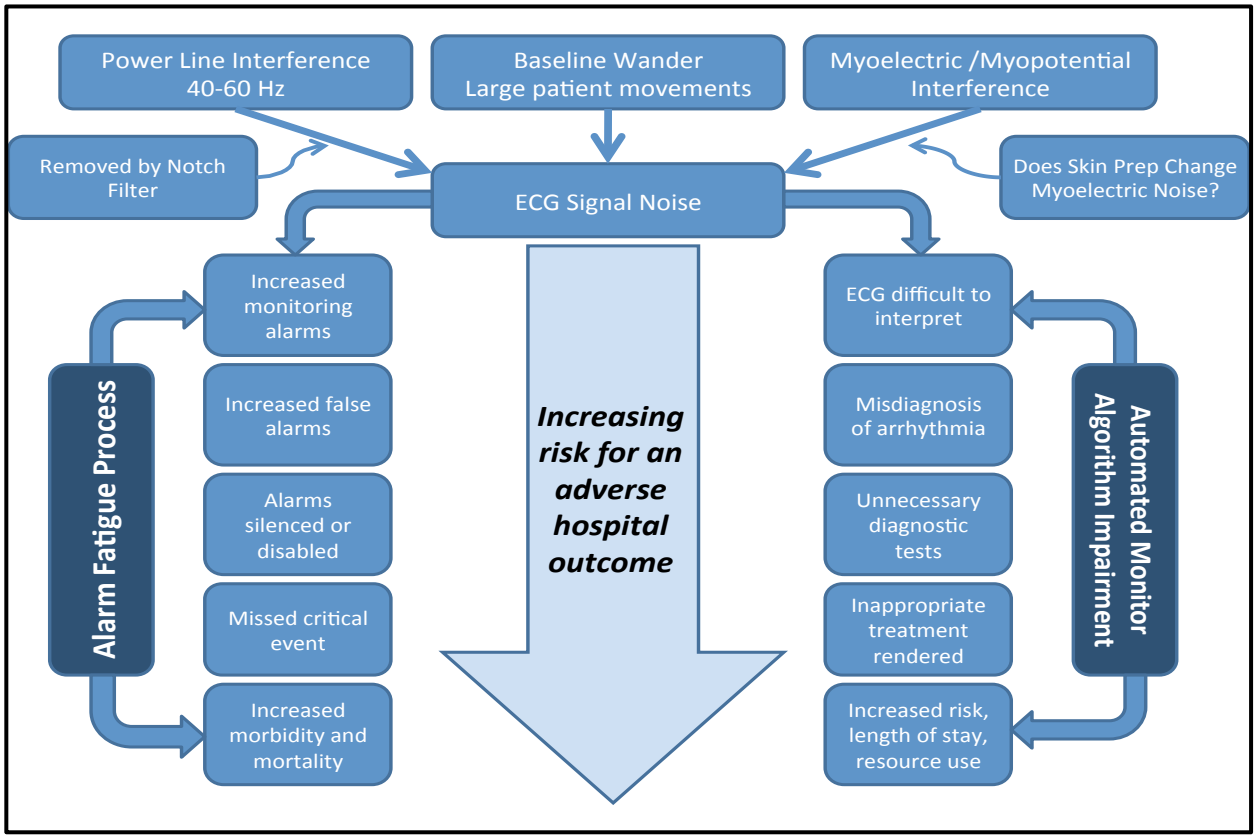
The lower arm of the original Fidler-Drew Model focuses on human over-reading of noisy

ECG signals. In this flow diagram, a noisy ECG signal makes human over-read difficult, impossible, or misleading. As an example, if an arrhythmia is misdiagnosed, the clinician may choose to order laboratory tests, echocardiograms, and imaging studies. These misinterpreted arrhythmias may also result in unnecessary hospital procedures such as cardiac catheterization, pacemaker implantation, or radiofrequency ablation (Knight et al., 1999). Every time a patient undergoes an invasive procedure, there is a risk of complications or death (Houtman et al., 2006; Knight et al., 1999, 2001; Vereckei, 2004). In addition to the clinical consequences of increased risk exposure for the patients, there are financial and ethical considerations. Financially, errors in continuous ECG interpretation result in unnecessary utilization of ancillary hospital services, and consultation with other healthcare providers. Ethically, this becomes an issue when invasive testing or device implantation occurs unnecessarily, and when the patient believes that they have a life-threatening health concern when they truly do not.

Revised Fidler-Drew Model of the Impact of ECG Noise

Although the original Fidler-Drew Model captured many of the salient components in the phases of both alarm fatigue generation and inappropriate testing and treatment of patients, it did not address outside influences at the source of the ECG signal. To incorporate a more comprehensive conceptualization of these processes, the model was further developed to better describe the sources of noise typically found in the ECG signal as shown in **Figure-2.8**.

Figure-2.8-Revised Fidler-Drew Model of the Impact of ECG Noise



There are three primary types of noise that corrupt the ECG signal. These are powerline interference from electric motors near the ECG monitor, baseline wander caused by low frequency but high amplitude movements of the patient such as rolling over and breathing, and myoelectric noise from muscle tissue underlying the electrodes. Of these three primary sources of ECG signal noise, the powerline interference is the simplest to eliminate from the signal. Since most electric motors operate on AC power, and most AC power in the United States is 60 Hz in frequency, a filter placed on the monitor processor removes all signals in the frequency range from 50-60 Hz. Although this filter removes a significant portion of the powerline interference, it does remove any signals from the patient that occur in that frequency band as

well. Fortunately, there are very few (if any) clinically significant findings on the ECG in that frequency range.

Baseline wander on the ECG is very difficult to remove, albeit easily identified by the large, rolling nature of an undulating baseline on the ECG tracing. Although it is a low frequency signal, the high amplitude does make interpretation of physiologic signals during baseline wander difficult or sometimes nearly impossible. The amplitudes achieved on the ECG by rolling a patient onto their side are large enough to obliterate the QRS of the ECG rending automated and human analysis of rhythm useless. Most sources of baseline wander are short-lived and do not cause significant intervals of time where the patient is unmonitored. Monitoring alarms from baseline wander are typically predictable, since staff manipulation is one of the primary sources (Atzema et al., 2006; Tsien & Fackler, 1997; Welinder et al., 2010).

Myoelectric noise in the ECG waveform is an expected source of noise in almost all patients. Patients in the intensive care unit who are receiving paralytic agents and are completely motionless may have no myoelectric noise in their ECG signals, but these subjects are the exception. Both the Mason-Likar and Lund electrode configurations place electrodes over muscle in most sites; so underlying myoelectric signals are destined to be received as noise in the signal while monitoring for myocardial electrical potentials at the skin. If the electrocardiographic potentials are greater in voltage than the myoelectric noise, then the tracing should still have some resemblance to an ECG tracing. It is known that in the face of myoelectric noise, both human and automated analysis is impaired, but the agreement between humans and computers still remains good. Farrell, et al. described a concordance rate between automated and human analysis of 89% even when there were high levels of myoelectric noise in the signal (R. M. Y. Farrell, B.J., 2004).

Although it is conventional thinking that noise in the ECG signal is a source of increased numbers of monitoring alarms, and particularly a high false alarm rate, research in this area is limited. Data from standard 12-lead ECG interpretation would suggest that the automated

algorithm would only make errors in arrhythmia and ST-segment analysis 11% of the time even in the worst signal-to-noise ratio conditions (R. M. Y. Farrell, B.J., 2004). The Fidler-Drew Model of the Impact of ECG Noise is subject to change in the future depending on additional research required to ascertain whether ECG signal noise truly contributes to alarms and false alarm rates in continuous monitoring situations.

CHAPTER 3: RESEARCH DESIGN & METHODS

Introduction

Although there is a recognized problem with alarm fatigue, the factors contributing to this situation are neither well delineated nor understood. Correct electrode placement is of paramount importance in obtaining ECG waveforms that are useful for comparative analysis; however, it is has been demonstrated that this is not done well by clinical staff (Rajaganeshan et al., 2008; Wenger & Kligfield, 1996). Making the electrode-skin interface better improves reception of electrical potentials across the skin and has been studied; however, this aspect of ECG monitoring has never been quantified between currently used or proposed ECG electrode locations (Tam & Webster, 1977). Furthermore, many clinicians believe that false ECG monitoring alarms are directly related to poor signal quality, but this has never been demonstrated through rigorous research. For these reasons, the relationships among electrode placement, ECG signal quality, and false monitoring alarms are unclear.

Study Design

This is a prospective, descriptive study with matched-paired sampling of patients serving as their own control through attachment of a pair of high-resolution Holter monitors. One Holter monitor was placed in the Lund configuration and second Holter in the Mason-Likar configuration. The study Holter monitors were in addition to the hospital monitoring system that was used for clinical decision-making and diagnosis. This paired, time-synchronized Holter monitoring approach is the only research design that will permit direct comparison between the Mason-Likar and Lund configurations for signal quality and false arrhythmia alarms during identical patient conditions. This method of data collection with synchronized and simultaneous ECG recording devices has been successfully conducted before in several studies, making this approach reasonable and feasible (Drew, Koops, Adams, & Dower, 1994; Drew et al., 1999).

Setting and Sample

The proposed research was conducted at the University of California San Francisco Medical Center in one adult ICU (10 ICC) and one adult PCU (10 CVT) with a total target enrollment of 100 subjects over a 6 to12-month time interval. Formal approval through the Institutional Review Board (IRB) was obtained through the University of California San Francisco Committee on Human Research (CHR). The approved version of the application for study number 10-04962 that received expedited approval is attached as **Appendix 1-CHR/IRB Approval**. This particular PCU and ICU provide care for a diverse spectrum of patient ages, national origins, medical conditions, and acuity levels. The PCU has 50 telemetry-equipped beds, and the ICU has 14 beds, both with the same GE Healthcare monitoring system to provide an ample pool of subjects. The large pool of monitored patients made recruitment feasible and practical. This medical center was chosen because of its proximity, past support for the mentor's program of research, and diverse patient population, represented by 12.7% Asian American, 9.8% African-American, 4.6% Latino, 0.7% Native American, and 4.9% mixed race.

Inclusion Criteria:

1. Admission to either ICU or PCU anticipated staying for ≥ 24 hours.
2. Age: 18 years and older.
3. Patient reads and speaks English for informed consent.

Exclusion Criteria:

1. Expected death or discharge in less than 24 hours.
2. Age under 18 or greater than 99.
3. Inability to consent due to sedation or altered mental capacity.

Study Procedures

Data collection began in November 2012 with a recurring weekly schedule. Tuesdays

and Saturdays were days on which subjects were approached for enrollment, and Wednesdays and Sundays were days for scheduled study monitor removals. The Drew ECG Monitoring research lab owns 18 Mortara H12+® Holter monitors which allowed for up to 9 patients per day to be enrolled. The study Holter monitoring systems were attached to the patient in addition to the standard hospital monitoring equipment used to guide clinical care. The Holter monitor data was collected simultaneously from both the Mason-Likar and the Lund electrode configurations, and this data was used for offline research analysis. These Holter monitors were chosen for this study since they were already owned by the research team, they are small and lightweight to minimize patient inconvenience, and they have a high-resolution sampling rate of 1000 samples per second (Hz) to produce technically adequate raw ECG signals.

The research team created enrollments packs that contained all necessary items including the Holter monitors, electrode lead wires, sealed packs of Philips wet-prep electrodes, skin preparation paper, Data Collection and Consent forms, as well as nursing instructions for the head of the bed. On data collection days, the research team met with the charge nurses for each unit to determine which patients would be included or excluded from enrollment due to anticipated death or discharge, pending major tests or surgery, or inability to communicate in English. All eligible patients were approached with a standard script to offer participation in ECG research. The consent form used in this study is shown in **Appendix-2-Consent Form.**

After obtaining informed consent, each subject was assigned a coded subject identifier and a study patient number. Each Holter was programmed to have the same time and date, as well as the coded 6-character subject identifier. The coded identifier followed this format:

- Character 1—either M or L to designate Mason-Likar or Lund configuration.
- Character 2—either 0 for no skin preparation, 1 for skin preparation.
- Character 3-5—a three-digit subject number 001 through 100.
- Character 6—0 designates a complete 24-hour recording, 1 designates a partial

recording interval.

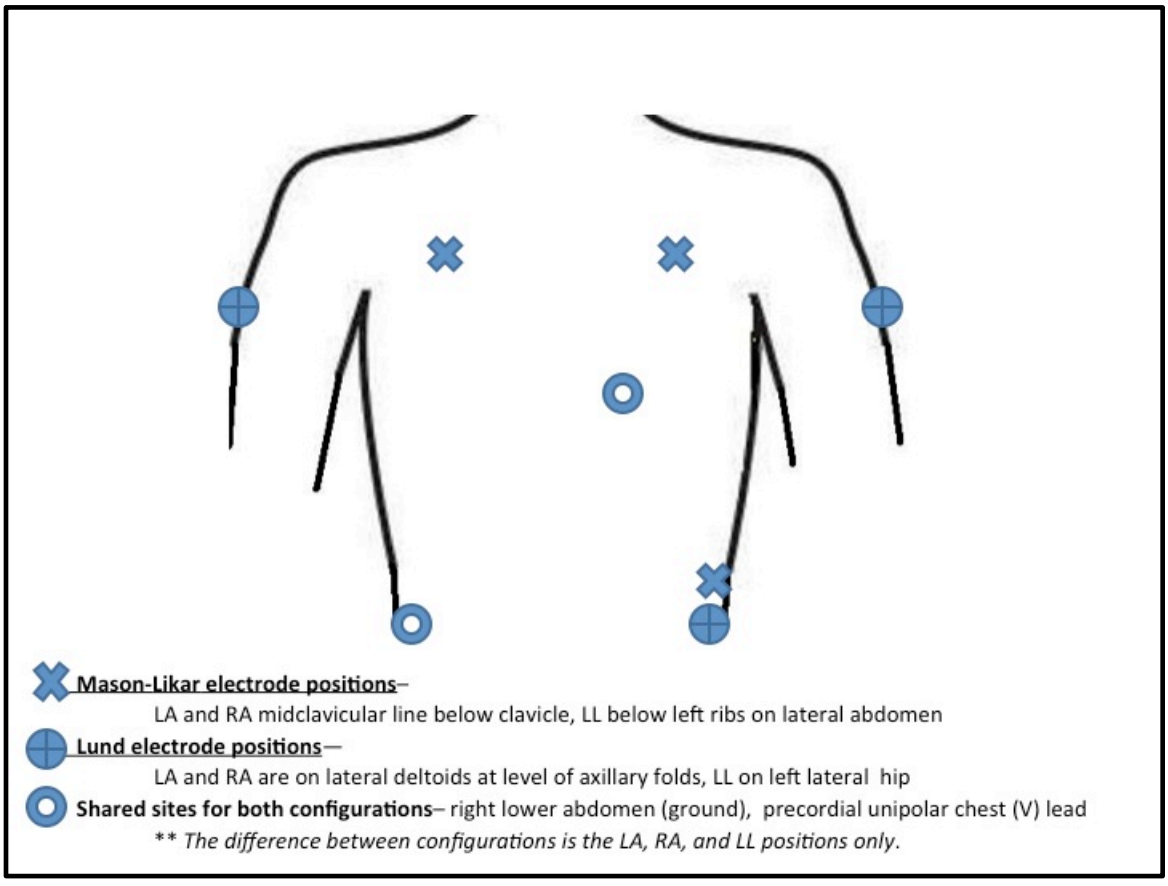
- Example M10341 is a Holter monitor connected to the Mason-Likar configuration to a patient receiving a skin preparation on subject 34 who did not complete the 24-hour data collection period.

Information was collected about the subjects in a standardized fashion using a preprinted Data Collection Form included as **Appedix-3-Data Collection Form**. The data gathered for each patient included demographics, unit at time of enrollment (ICU or PCU) and study completion, vital signs, diagnoses, medications, laboratory values, ECG and cardiac diagnostic testing results, and mobility status for later sub analyses. Each enrolled subject had a pair of lightweight Mortara H12+® Holter monitors attached by study staff, Richard Fidler, with verification by registered nurse research assistants. Verification included patient consent, skin examination, verification of skin preparation randomization and performance, accuracy of each electrode placed in each configuration, and accuracy of the Holter to the correct configuration. The clocks in both Holter monitors were synchronized so that direct comparisons of ECG events were made between the two lead configurations under investigation.

For each subject, one Holter monitor was attached to six electrodes placed in the proscribed Mason-Likar configuration including right arm electrode in the right infraclavicular fossa, left arm electrode in the left infraclavicular fossa, ground electrode on the right lateral chest wall, left leg electrode placed on the left lateral abdomen below the final rib but above the iliac crest, V1 in the 4th intercostal space along the right sterna border, and V5 place in the anterior axillary line in the 5th intercostal space between the standard V4 and V6 positions. The second Holter monitor was attached to six electrodes placed in the proscribed Lund configuration placing the left arm electrode on the outer, lateral left deltoid below the level of the axillary fold, the right arm electrode placed on the right lateral deltoid below the level of the axillary fold, the ground electrode (reference electrode) placed on the right lateral chest wall, the left leg electrode placed on the left lateral hip, V1 in the 4th intercostal space along the right

sterna border, and V5 place in the anterior axillary line in the 5th intercostal space between the standard V4 and V6 positions. These electrode configurations are shown in **Figure-3.1. F**oam wet-prep electrode (Philips, Model 40493-D) were used in this study due to their high-quality adhesive ring, and proposed superior signal quality. If the patient was randomized to an abrasive skin preparation as part of participation in this study, a finger-shaped adhesive-backed fine grit sandpaper (Philips product number M4606-A) was used with three strokes at each electrode site in both electrode configurations. The researchers made every attempt to apply the same amount of pressure on the skin preparation sites each time in every subject.

Figure-3.1--Mason-Likar compared to Lund Electrode Configurations



After the Holter monitors were connected to the subject, a marking pen dedicated to each subject was used to draw a circumference around each electrode to facilitate replacement

of any electrode if an electrode became dislodged. The electrodes were dated, timed, and initialed by the study staff at the time of study monitor attachment. To reduce subject burden for participation in the study, a neck pendant type of pouch was provided to hold both Holter monitors to facilitate mobility. Over the 24-hour monitoring period, research staff would visit enrolled subjects to verify that electrodes were still on the skin, lead wires were connected to the electrodes, Holter monitors were still connected to the correct electrodes, and that the Holter monitors were still recording. Enrollment and initial connection typically occurred between 9am and 2pm, with subsequent visits at 5-6pm, 10-11pm, 9-10am, and removal at the appropriate time. If the patient elected to withdraw from the study at any time, any reason cited was noted on the Data Collection Form.

Prior to removal of the Holter monitors, each subject was asked whether any electrodes were troublesome. Subjects were asked to point directly to any electrode(s) on their body that was problematic, and then asked to comment on the nature of the problem the electrode created such as (1) limitations to movement, (2) discomfort at the electrode site, and (3) impact on sleep due to the electrode location. On a Likert scale from 0-10, the subject rated any electrode site that was deemed problematic by the patient for any reason. If patients volunteered additional comments about electrode sites, the monitors, lead wires, and the experience of being monitored, this qualitative data was also collected. This subjective subject data was entered onto the data collection form by study staff. Immediately after removal, all Holter monitoring equipment was cleaned following standard hospital protocol. Data extraction from the Holter monitors occurred in the Drew ECG Monitoring Research Lab where the data was transferred and stored electronically on a secure server for offline and offsite blinded computerized analysis.

Instruments and data collection

Missing baseline data or follow up data was not anticipated; however, some subjects

were not be able to provide all 24 hours of monitoring data due to unanticipated discharge from the hospital, panic attacks, and perceived bother. In these cases, the researchers analyzed the number of hours of continuous monitoring data that has been acquired as long as both Mason-Likar and Lund ECG information was available for direct comparison. There were multiple types of data generated through the aforementioned collection, and the processing of this data will be described in detail. The raw ECG signal sampled at 1000 Hz provided adequate signal resolution for many analyses. ECG data was redundantly saved onto 2 encrypted hard disc drives as well as a secure cloud server in the proprietary Mortara HScribe® format. Files were also converted files into binary files for universal transfer and processing, and data collection form hard copies were in a locked filing cabinet in the security controlled Drew ECG Monitoring research laboratory.

Data Processing

After the data for all 100 subjects was gathered, Dan Schindler of the Drew ECG Monitoring research laboratory, who was not involved in the study, was tasked with blinding study staff and consultants to the raw data files. Using Matlab®, a random number assignment routine was used to code subject identifiers for each of their Mason-Likar and Lund Holter monitor files. All patients had at least two files, one for the Mason-Likar and another for the Lund electrode configurations. Because the Holter monitors save each 24-hour period of time as a file, some subjects that provided more than 24-hours of monitoring data had a full 24-hour file and a second file for the hours beyond the 24-hours in each monitoring configuration.

Using Matlab®, each of the ECG data files underwent conversion from raw analog ECG data to a binary file format. The binary file format was chosen as the common format that could be used for each of the quantification methods that will be described below. The coded binary data files were saved onto three different Western Digital 1.0TB (terabyte) hard disc drives. One of these drives was created as a backup copy of the binary data, and the other two were mailed

to each consultant that performed the offsite, offline, blinded analysis for ECG signal quality and ECG arrhythmia alarms.

Noise Assessment and Measurement Using Hook-Up Advisor®

Noise in the ECG signal is proposed to be a fundamental problem associated with both misdiagnosis and increased false arrhythmia alarms. It is crucial that ECG noise be evaluated scientifically, objectively, and accurately quantified. ECG data from both the Mason-Likar and Lund Holter monitors was analyzed for ECG signal quality in a blinded manner using a standardized, FDA-approved computerized algorithm. The Hook-Up Advisor® algorithm (FDA approval number K042177) was developed by GE Healthcare for use in the MAC 5000 electrocardiograph. The algorithm analyzes up to 500 samples per second, which is higher than the 125 Hz sampling rate of the monitoring system currently used at the proposed site of research, but also less than the study Mortara H12+ ®Holter monitors that obtain ECG signals at 1000 Hz. To be able to make a directly analogous comparison of the research ECG data to the typical hospital monitoring system, Matlab® was used to downsample the study ECG samples from 1000 Hz to 125 Hz consistent with the hospital equipment. These downsampled signals could also be processed with Hook-Up Advisor®.

Standard monitoring equipment uses filters to remove unwanted frequency bands from the ECG signals and waveforms. Power line or alternating current (AC) interference is typically eliminated using a notch filter that has a single preset frequency of 40 Hz to 60 Hz to remove unwanted interference from the ECG signal due to electrical sources. In addition to the notch filter, Hook-Up Advisor® has automated adaptive filtering in the 40 to 60 Hz bandwidth identical to the hospital monitoring system to remove much of the power line or AC interference noise from sources of electrical static, such as electric motors. The Root Mean Square (RMS) evaluation is the standard practice used by monitoring companies and the U.S. Food & Drug Administration to assess for ECG noise. The computer algorithm follows the principles of

evaluating the signal through this RMS process to create the baseline for subsequent comparison (Batchvarov, Hnatkova, & Malik, 2002; Cherkassky & Kilts, 2001; Farrell & Rowlandson, 2006).

Electrocardiographic noise is determined by computerized assessment of QRS complexes for false deflections. The algorithm examines the lead energy content of each QRS for amplitude in the context of R-R intervals. Hook-Up Advisor® then proceeds to assess and measure the other two types of noise in the ECG signal, myoelectric noise and baseline wander. Myoelectric noise is detected by Hook-Up Advisor® by counting the number of deflections exceeding a preset limit in any one-second time interval. To match this particular hospital's monitoring system's version, Hook-Up Advisor used the threshold of 40-45 deflections per second across the horizontal baseline in any lead or leads of the ECG signal as the definition for the presence of myoelectric noise (Christov & Daskalov, 1999; Farrell & Rowlandson, 2006).

Baseline wander is the second type of noise that interferes with ECG signal smoothness. Baseline wander is the low frequency, high amplitude changes that occur with large patient movements, such as when patients roll over; make slow movements while stretching, or even movement of the chest with the respiratory cycle. This type of noise manifests on the ECG waveform as an undulating baseline. When the ECG baseline wanders, it is impossible to establish the true isoelectric T-P and P-R segments of the cardiac cycle that are used to assess whether there is ST-segment elevation or depression in patients experiencing acute myocardial ischemia. The computerized noise algorithm further characterizes the baseline wander as either (1) sway or (2) saturation. Sway is determined through the low-pass filtering system by determining the difference between the minimum and maximum of the signal. If the difference exceeds the proprietary-set threshold, the algorithm annotates the signal as having sway. Saturation of the ECG signal is defined by Hook-Up Advisor® as ECG signals that exceed +/- 4.8 millivolts amplitude for greater than 100 continuous milliseconds (Farrell & Rowlandson, 2006). Once the signal is thoroughly analyzed by Hook-Up Advisor® for power line

interference, myoelectric noise, and baseline wander, the computerized noise algorithm assigns a color code to the quality of the signal from each ECG lead. Signal quality in limb leads I and II and the unipolar chest leads (V leads) are designated as “green” for an ECG signal that has an acceptable level of noise and is considered normal. When the signal is lost or becomes so distorted that it is completely unanalyzable, the algorithm assigns that signal to the color “red”. For ECG noise that is above the green but below the red thresholds, a designation of “yellow” is given to the signal (Farrell & Rowlandson, 2006). Thresholds are proprietary information and not disclosed to the research team; however, the thresholds are the same across all GE Healthcare ECG devices.

Data were delivered to Dr. Robert Farrell, PhD, at GE Healthcare ECG department on a Western Digital 1.0 TB disc with blinded file numbers, and each file was in binary format. Using Matlab®, Dr. Farrell was able to reconstruct the waveforms for analysis using Hook-Up Advisor®. Each file was run through the Hook-Up Advisor® that was able to produce two types of data output for analysis. First, Hook-Up Advisor® produced a Microsoft Excel™ spreadsheet for each subject. Each one of these spreadsheets provided an hourly summary of signal quality for Leads I, II, V1, and V5. This hourly summary of time spent in each color-coded signal quality state was derived by analyzing the 360 10-second segments for each hour of monitoring as shown in **Appendix-4-Summary of Hook-Up Advisor Signal Quality Statistics**. A second form of data output was also generated for each file that provided a deeper analysis of each 10-second segment broken into 2.5 second intervals where powerline interference, baseline wander, and myoelectric noise were quantified with the corresponding color-code for signal quality as shown in **Appendix-5- Sample Hook-Up Advisor® Events Output.**

Since Hook-Up Advisor® was designed and used for evaluation of standard diagnostic 12-lead ECG’s and not for evaluating signal quality in a continuous monitoring format, this represents an adaptation of this program for a novel use of an approved technology.

Arrhythmia Assessment and Measurement

Although there are many alarms built into monitoring systems for arrhythmia detection, we focused on false alarms that would, if misinterpreted, signal a life-threatening arrhythmia event. Since accurate true positive arrhythmia alarms are the desired effect of monitoring, the component of false alarm analysis assisted with answering Specific Aim # 2, to determine whether there is a difference between the Mason-Likar and Lund electrode configurations in the number and type of false lethal arrhythmia alarms. In particular, this study examined the rate of false alarms for asystole, extreme bradycardia, ventricular tachycardia, and ventricular fibrillation. The computerized algorithm for arrhythmia analysis is challenged in making the correct rhythm determination when noise is present in the ECG signal. For this reason, the evaluation of noise and false monitoring alarms are inextricably related.

Much like human ECG analysis, the computerized algorithm for arrhythmia detection begins with R-wave detection first to compute the most easily derived parameter of heart rate. Absence of any R-waves for greater than 6 seconds meets technical criteria for asystole. For bradycardia determination, the algorithm determines that the R-wave frequency represents a heart rate below 50 beats per minute as technically significant bradycardia. For asystole and bradycardia determination, it is apparent that R-wave detection be accurate to prevent false asystole and bradycardia alarms. For ventricular tachycardia, the arrhythmia algorithm must first detect R-waves, and then measure the width of the QRS complexes for each R-wave. To meet the algorithmic criteria for ventricular tachycardia, there must be 6 or more sequential QRS complexes that exceed 0.12 seconds in duration and occur at a frequency greater than 100 per minute. To determine whether ventricular fibrillation is present, the arrhythmia algorithm must identify either an absence of QRS complexes with high-frequency fibrillatory waves or the presences of R-waves occurring at greater than 300 per minute.

Operational Definitions of Lethal Arrhythmia Alarms

For the purposes of this study, it is crucial to understand the definitions of each lethal arrhythmia alarm triggered by the arrhythmia identification algorithm. In this study, lethal arrhythmia alarms are defined as asystole, bradycardia, ventricular tachycardia, and ventricular fibrillation each defined as below consistent with the monitoring systems used in the study ICU and PCU:

1. Asystole is the absence of any R-waves for greater than 6 seconds.
2. Bradycardia is a continuous period of 5 to 6 seconds where the heart rate is computed to be less than 50 beats per minute.
3. Ventricular Tachycardia is six or more beats that have a QRS width greater than 0.12 seconds at a rate greater than 100 beats per minute.
4. Ventricular Fibrillation is rhythm that has a heart rate measured at greater than 300 beats per minute.

Measurement of Lethal Arrhythmia Alarms

Following a similar procedure to the signal quality analysis using Hook-Up Advisor®, identification of lethal arrhythmia alarms was accomplished by offsite, offline, blinded analysis of ECG waveforms obtained during this study. A Western-Digital 1.0TB hard drive was mailed to Dr. Mikko Kaski, PhD, of GE Healthcare in Helsinki, Finland who ran the data through EK-Pro® by GE Healthcare. Identical data was sent to Dr. Farrell who used Hook-Up Advisor® by GE Healthcare. Using Matlab® to convert the binary files into ECG signals that were then run through both algorithms, Drs. Kaski and Farrell used their respective analysis algorithms on the blinded raw data.

For appropriate analysis that would replicate the alarm conditions that would have been triggered on the hospital units, Dr. Kaski was provided minimal clinical information regarding cardiac devices to create appropriate settings in EK-Pro®. For recognition of pacemakers and

other intracardiac devices, a sampling rate of 125 Hz is inadequate. Dr. Kaski was made aware of any file that had any of the following devices were present: (1) temporary pacing transvenous wire, (2) permanent pacemaker implant, (3) implantable cardioverter-defibrillator, and (4) left-ventricular assist device. In the hospital monitoring system, if the "pacing detection" feature is activated, the sampling rate is changed from 125 to 250 Hz to detect the very brief electronic pacemaker spike. In the case files where devices were present downsampling the signals to 250 Hz from the original 1000 Hz samples was performed to match the hospital monitoring system.

After running all of the study data through EK-Pro®, data were sent back to UCSF for adjudication of the alarm conditions. Co-investigator, Richard Fidler, took primary responsibility in monitoring alarm adjudication. Using Matlab®, each alarm condition identified by EK-Pro® was presented in a Matlab® waveform viewing program that was designed for the purposes of this study. A graphical user interface requested user input to assign “true”, “false”, or “unknown” to the alarm condition and the waveform present. If the arrhythmia identified by EK-Pro® did not meet the operational definition of that arrhythmia, the alarm condition was labeled as “false”. If criteria were met and a valid alarm condition was identified, it was labeled as “true”. Any algorithm identified alarm that was not clearly meeting or not meeting operational definition criteria was escalated to Dr. Barbara Drew, PhD, RN, who is internationally known for her work in arrhythmia recognition.

As a part of the Matlab® viewing routine used for alarm review, each reviewed alarm for each data file was placed into a .fp file for false positive alarms, .tp files for true positives, and .un files if adjudicated as unknown. Prior to unblinding the source information for that file, co-investigator Richard Fidler reviewed the alarms. The alarms analysis was conducted without knowledge of that file's electrode configuration source or any patient factors.

Alarms files were tabulated and placed into a spreadsheet using Microsoft Excel™. The alarms spreadsheets were constructed using file ID’s, not patient identifiers. This spreadsheet

was unblinded by Dan Schindler of the Drew ECG Monitoring Research laboratory and returned to the research investigators.

Patient Factors Assessment for ECG Monitoring

Although the original intent of this aim was to determine the patient perceptions of wearing electrodes on the outer arms instead of the chest, this qualitative data assessment may influence recommendations for ECG monitoring. It would not matter which lead configuration provided less ECG noise and fewer false alarms if that configuration resulted in patient discomfort, interruption to sleep, skin breakdown, or significant limitation in movement. For this reason, the research team devised a data collection form that will allow communicative patients the option of sharing their perceptions of the ECG monitoring electrode sites, as well as any other comments about the monitoring equipment or experience. Subjects were able to first identify then rate any individual electrode site(s) that were troublesome and rate them on a Likert scale 0-10. If they had identified any troublesome electrode sites, the study staff ascertained the reason. Additionally, study staff recorded any observed skin irritation from any electrode site on the data collection form.

Data Analysis Plan for each Specific Aim

Aim 1

In patients in the ICU or PCU, determine whether the Mason-Likar and Lund electrode configurations differ in the amount of myoelectric noise and baseline wander produced in the ECG recordings over a 24-hour monitoring period.

The data analysis for ECG noise included the following variables: (1) time spent with “red” signal quality, time spent with a “yellow” signal quality, and time spent in a “green” signal quality. Since no prior research has described typical longitudinal signal quality data over a 24-hour time

period, descriptive statistics on the aggregated data from both electrode configurations is warranted. Since simultaneous data from Holter monitors in both the Mason-Likar and Lund electrode configurations were collected, we are able to do a direct comparison of the continuous variables of signal quality for each electrode configuration. Data for signal quality was normally distributed and allowed for parametric statistical approaches. The matched paired t-test is the most appropriate statistical test for the comparison of the means for each of these variables.

Further analysis included comparison of the signal quality between ICU and PCU patients, with a comparison of the Mason-Likar to Lund electrode configurations depending on the ambulatory status of the patients, and the type of unit. More analyses that were conducted based on skin preparation status will be discussed in Aim 3. Additionally, further analysis by Pearson's R correlations with the number of ECG alarms and signal quality to determine whether any relationship between ECG signal quality and false lethal arrhythmia alarms exists.

Aim 2

In patients in the ICU or PCU, determine whether the Mason-Likar and Lund electrode configurations differ in the number of total alarms and false life threatening alarms generated by the monitor system's arrhythmia analysis algorithm over a 24-hour monitoring period.

The data analysis for the false monitoring alarms aim of this research included the following variables: total number of true lethal arrhythmia alarms with each of the two lead configurations, the number of false positive ventricular tachycardia, ventricular fibrillation, extreme bradycardia, and asystole arrhythmia alarms for the Mason-Likar and Lund electrode configurations. The researchers over-read each life threatening alarm condition by two methods, analyzing ECG waveforms using the Matlab® viewer, as well as using Mortara HScribe® Holter monitor software. Other independent variables in the analysis of alarms are (1) the total number

of arrhythmia alarms triggered in the 24-hour monitoring period, (2) breakdown of alarms by subject, (3) the number of true and false alarms generated over the monitoring period as a proportion of the total alarms triggered, and (4) Pearson's R correlation of monitoring alarms with the amount of ECG noise during time interval analysis. The initial statistical analysis plan for lethal arrhythmia alarms data was planned to be the matched paired t-tests to compare the mean numbers of alarms broken down by unit type, ambulatory status, and demographic factors.

Unfortunately, the arrhythmia alarms data is not normally distributed, being over-dispersed and highly skewed. Initial attempts at statistical analysis prompted further consultation with a statistician, Dr. Steven Paul, PhD, of the University Of California San Francisco School Of Nursing. Limited options existed for the primary statistical analysis of this arrhythmia alarms data, and Related-Samples Wilcoxon Signed Ranks test was conducted. Frequency data was also tabulated confirming that the majority of false lethal arrhythmia alarms were generated by a small subset of the sample.

Because the data for false lethal arrhythmia alarms lacked the robustness to provide adequate information in the context of signal quality, further consultation was sought with Dr. Xiao Hu, PhD who has extensive experience in the analysis of monitoring alarms. Based on the recommendations of Dr. Hu and post-doctoral fellow Dr. Rebeca Salas-Boni, PhD, more analysis was recommended and conducted on the original data.

Due to the novelty of this study, comparison of these results to other research reports was not available. Improving the quality of the data analysis required the review of all of the patients that had any monitoring alarms in the database. Beginning with the exclusion of any leads off conditions, all of the alarm conditions were re-examined in the Matlab® viewer, with a subsequent review of the time-stamped HScribe® Holter monitor waveforms to validate the waveform in question. Any alarm condition with a lead-off condition in any of the leads was excluded from the final phase of analysis.

Since one of the major aims of this study is to determine whether differences exist between the Mason-Likar and Lund electrode configurations in signal quality and false lethal arrhythmia alarms, a reanalysis and recount of all monitoring alarms was performed with the added depth of time-synchronization. To perform the McNemar test to compare differences, the assumptions of a paired sample was met; however, it requires the metrics of which alarms occurred in the Mason-Likar only, the Lund only, and both the Mason-Likar and Lund electrode configurations. A Matlab® routine was written specifically for this purpose by Dr. Salas-Boni to sort the alarms by time and electrode configuration.

Determining the relationship between signal quality and lethal arrhythmia alarms was accomplished as follows. For each lethal arrhythmia alarm, the signal quality designation of green-yellow-red was determined. Each type of alarm was then analyzed for the number of alarms in each color-coded signal quality state. To normalize this data for each subject, the number of each alarm in each signal quality was divided by the time spent in each signal quality state. This approached allowed the use of a match paired t-test to compare the normalized values for each alarm type in each signal quality state as a proportion of the total amount time spent in each signal quality state.

Aim 3

In patients in the ICU or PCU, determine whether an abrasive skin preparation has an impact on ECG signal quality or false lethal ECG arrhythmia alarms in either the Mason-Likar or Lund electrode configurations over a 24-hour monitoring period.

In this sample of 100 patients undergoing randomization to either receive standard electrode placement or an abrasive skin preparation, only one patient that was randomized to the skin preparation declined to participate in that component of the research. Many clinicians believe that false arrhythmia alarms are rooted in poor signal quality, and manufacturers

recommend a skin preparation prior to electrode placement. Controversy exists over what, if any, skin preparation should be used. This study examined the use of an abrasive skin preparation as an intervention strategy to improve signal quality and decrease false lethal arrhythmia alarms.

The analysis for this aim began with conducting a t-test between the Mason-Likar and Lund electrode configurations for signal quality as measured by the mean percent of time spent in green, yellow, and red signal quality. To determine whether the Mason-Likar and Lund electrode configurations were equally impacted by the use of the abrasive skin preparation, a comparison of the difference of the means between skin prepared and standard electrode application was conducted. The data analysis plan for Aim 2 will also satisfy the question with regard to skin preparation and arrhythmia alarms.

Aim 4

Describe patient perceptions of each electrode site with regard to comfort, interruption to sleep, and impairment of movement, as well as describe the unsolicited patient perceptions of monitoring equipment and the experience of being monitored.

Of the proposed aims of this study, this secondary aim presented challenges because of subjective data collection and analysis. This portion of the study was not expected to include all subjects, since there was a portion of the sample that was noncommunicative due to disease, intubation, or sedation. It was anticipated that subjects for this aim were more likely to be PCU patients that were being mobilized in preparation for hospital discharge or have less acute illness. The data on patient factors was only be able to be performed with subjects that could answer the questions independently, without surrogates providing answers. Data analysis included descriptive statistics only on proportions of the sample that offered this type of information, and this data will be placed in a table for better organization.

Power Analysis

Enrolling 100 subjects over a 6-month time period was expected to be feasible, and provided about 2,400 hours of ECG data for each of the two lead configurations, far exceeding the amount of data collected in any prior research conducted to date. Subject recruitment was not problematic, although there were challenges with early discharges, patient declination, language barriers, and the lack of capacity to provide informed consent. There were large numbers of noisy periods and false alarms to analyze. Statistical consultation was obtained with Steven Paul, PhD, who is a staff biostatistician at the University of California San Francisco School of Nursing. A power analysis calculation was conducted using nQuery® for both of the specific aims of this study. Since no prior studies have been conducted to establish an effect size, the remaining components of the power analysis were employed to determine the effect size. A 2-tailed approach with an alpha level of 0.05 at a power level of 0.80, using a sample size of 100 subjects yielded a small effect size with d=0.283. This power analysis demonstrates that with 100 subjects each providing 24 hours of data through two separate Holter monitors, we have the ability to detect a difference in electrode configurations as small as 0.283 standard deviation units.

Chapter 4. RESULTS

Study Overview

This chapter will present the findings of the primary analysis of original data conducted for a dissertation research study. The results will be presented by specific aims and hypotheses. The initial research questions that framed this study were to determine whether two commonly used electrode configurations and skin preparation had an effect on ECG signal quality, monitoring alarms, and patient comfort.

Data analysis was conducted using PASW (SPSS) Statistics Version 18 (IBM) in conjunction with statistical consultation with Dr. Steven Paul. Dr. Paul has reviewed all results for statistical accuracy, and the co-principal investigators on this study have verified the context of the statistical analyses to ensure clinical relevance. Dr. Rebeca Salas-Boni, PhD, who has experience in analyzing data pertaining to monitoring alarms, performed additional analysis and results verification. Signal quality data was analyzed in a blinded fashion by Dr. Robert Farrell, PhD of GE Healthcare (Milwaukee, Wisconsin) using Hook-Up Advisor® (GE Healthcare) to analyze the data in 10-second segments. Blinded analysis using EK-Pro® for arrhythmia recognition was conducted by Dr. Mikko Kaski, PhD at GE Healthcare in Helsinki, Finland.

As discussed in prior chapters, this study had four specific aims, three of which are suited to hypothesis testing, and the final aim is a mixed method qualitative-quantitative aim. The results of each aim will be presented separately, with a synthesized contextualization in Chapter 5- Conclusions and Discussion.

Aim 1: *In patients in the ICU or PCU, determine whether the Mason-Likar and Lund electrode configurations differ in the amount of myoelectric noise and baseline wander produced in the ECG recordings over a 24-hour monitoring period.*

Aim 2: *In patients in the ICU or PCU, determine whether the Mason-Likar and Lund*

electrode configurations differ in the number of total alarms and false life threatening alarms generated by the monitor system's arrhythmia analysis algorithm over a 24-hour monitoring period.

Aim 3: *In patients in the ICU or PCU, determine whether an abrasive skin preparation impacts the ECG signal quality and number of false life threatening alarms generated by the monitor system's arrhythmia analysis algorithm over a 24-hour monitoring period.*

Aim 4: *Describe patient perceptions of each electrode site with regard to comfort, interruption to sleep, and impairment of movement, as well as general perceptions of the experience of being monitored.*

Sample Results and Description

Sample

This study was conducted at a single university medical center in San Francisco, CA using a convenience sample of voluntary subjects from cardiac intensive care and progressive care units. Data was gathered between November 4, 2011 and June 2, 2012 with 687 patients approached to achieve a total enrollment of 100 volunteer subjects. The sample was comprised of 19 subjects from the ICU and 81 subjects from the PCU, with 20 subjects self-described or ordered bedrest and the remaining 80 ambulatory. Three subjects had LVAD (left-ventricular assist devices) and 10 subjects had pacemakers or defibrillator/pacemakers. Descriptive characteristics of the sample are seen in **Tables-4.1 and 4.2**.

Table-4.1—Subject Characteristics

	Mean	Min-Max (Range)	Standard Deviation
Age (years)	58.9	22-95 (73)	16.4
Height (Gamble et al.)	171.0	144.8-198.1 (53.3)	11.4
Weight (kg)	81.8	38.4-165.0 (126.6)	23.7
BMI	27.9	14.1-47.4 (33.3)	7.5
Gender	Male = 55	Female = 45	

Table-4.2--Subject Ethnicities

	Female (n=45)	Male (n=55)
Latin	2 (4.4%)	5 (9.1%)
Asian	4 (8.8%)	6 (10.9%)
African-American	6 (13.3%)	5 (9.1%)
Caucasian	28 (62.2%)	36 (65.5%)
Native American	0	1 (1.8%)
Other or Mixed	5 (11.1%)	2 (3.6%)

Results—General Data Collection

All 100 subjects provided ECG data for analysis in both the Mason-Likar and Lund electrode configurations. A total of 142,845 minutes (2380.75 hours) were gathered in each electrode configuration for a total of 4761.5 hours of analyzable data. The mean recording time

for each subject is 1428.45 minutes (23.8 hours), with a range of individual data contribution as brief as 86 minutes but as long as 1650 minutes (27.5 hours). For the subjects providing less than 24 hours of data, the primary reason for early discontinuation of study monitors was earlier-than-expected discharge from the hospital. There were 2 subjects that elected to withdraw at 8 hours, and 2 other subjects experienced panic attacks requiring removal of the study Holter monitors as well as the hospital monitoring devices.

Results—Presented by Specific Aim--ECG Electrode Configuration, Signal Quality, False Lethal Arrhythmia Alarms, and Patient Perceptions of Being Monitored

Aim 1: ***In patients in the ICU or PCU, determine whether the Mason-Likar and Lund electrode configurations differ in the amount of myoelectric noise and baseline wander produced in the ECG recordings over a 24-hour monitoring period.***

Hypothesis 1: There is no difference in ECG signal quality between the Mason-Likar and Lund electrode configurations.

ECG waveform data was analyzed for signal quality using Hook-Up Advisor® from GE Healthcare. Typically used for analyzing signal quality in diagnostic 10-second standard ECG's, Hook-Up Advisor® was adapted for use in the study by analyzing continuous monitoring data broken down into 10-second segments of time. To use all available data from all subjects, the analysis was performed on the percentage of total time for each patient spent in the three signal quality states. Analysis of data aggregated from both Mason-Likar and Lund electrode configurations is shown in **Table-4.3.** Data regarding signal quality is distributed normally and can undergo parametric analysis. This study showed that patients spent 62.5% of their time with

a clean, “green” signal quality level, with 4.2% of their time spent in an unreadable “red” condition. There were some subjects that spent considerable time in a “yellow” or “red” state, as shown by the range of the percent time spent in each level of signal quality in **Table-4.3**.

Table-4.3--Aggregated Mason-Likar and Lund Electrode Configurations Descriptive Statistics for the Entire Sample (n=100) Showing the Percentage of Time Spent in Each Level of Signal Quality

Level of Signal Quality	N	Minimum %	Maximum %	Mean	Standard Deviation
% Green Segments	100	5.3	96.1	62.5	18.9
% Yellow Segments	100	3.9	94.6	33.3	16.6
% Red Segments	100	.00	80.8	4.2	14.0

The Impact of Electrode Configuration on ECG Signal Quality

For all the subjects in the study, ECG signals from both the Mason-Likar and Lund electrode configurations were collected simultaneously in each subject. This research team used a null hypothesis approach to determine whether any difference exists in signal quality between the Mason-Likar and Lund electrode configurations based on the mean percentage of time each subject spent in “green”, “yellow”, or “red”. This study shows that on average, the signals from the Mason-Likar configuration provide a larger percentage of time in the best “green” signal quality than the same subjects in the Lund electrode configuration. The amount of time spent in a “yellow” or “red” signal quality state was greater for the Lund electrode configuration as shown in **Table-4.4**. In this case, we reject the null hypothesis since there is a significant difference in the mean amount of time subjects spend with “green”, “yellow”, or “red” signal quality between the two electrode configurations.

Table-4.4—ECG Signal Quality as Presented by Mean Percentage of Time Spent in Green-Yellow-Red Status for the Sample (N=100) by Electrode Configuration

	Mason-Likar (SD)	Lund (SD)	Mean Difference	p-value
Mean % Time in Green	66.6 (16.8)	58.4 (20.0)	8.2	<0.001
Mean % Time in Yellow	31.6 (15.6)	35.1 (17.4)	-3.5	0.011
Mean % Time in Red	1.8 (8.1)	6.5 (17.8)	-4.7	0.007

More Clean "Green" Signal Quality for Mason-Likar Configuration

The relationship between Mason-Likar and Lund electrode configurations regarding signal noise has not been clearly quantified prior to this study. The Mason-Likar configuration was found superior to the Lund, averaging 66.6% "green" segments while the Lund electrode configuration averaged 58.4% "green" segments per patient ($p<0.001$). The Mason-Likar electrode configuration provided a mean of 8.2% more "green" signal quality time than did the same subjects in the Lund. Using a 95% confidence interval, this mean difference is at least as much as 5.0% but could be as much as 11.5% more time spent in "green" by the Mason-Likar configuration.

More "Yellow" Signal Quality Time for Lund Configuration

The time spent in a "yellow" signal quality state represents suboptimal monitoring time, making it challenging for automated algorithm and human interpretation of ECG rhythm. During a "yellow" signal quality state, the automated algorithm reduces arrhythmia alarms for all but lethal arrhythmias, making time spent with a "yellow" signal quality a vulnerable time for

patients. Comparison of the Mason-Likar and Lund electrode configurations for the percentage of time spent with “yellow” signal quality is significantly different, with the Mason-Likar spending a mean of 31.5% of time in “yellow”, while those same subjects in the Lund configuration spent 35.1% of their time in “yellow”. The mean difference between the Mason-Likar and Lund configurations is 3.5% favoring the Mason Likar. Using a 95% confidence interval, this difference could be as little as 0.8% of the time, but may be as much as 6.3% more time spent in “yellow” status in the Lund configuration (p=0.011).

More “Red” Signal Quality Time for Lund Configuration

Although the time spent in a “yellow” status is considered suboptimal, the amount of time a patient spends with a signal quality classified as “red” is particularly troublesome. Time spent with a “red” signal status is time spent with arrhythmia alarms suspended, as the signal is nearly or completely uninterpretable. On average, patients in the Mason-Likar configuration spent 1.8% of the time in “red”, while the same subjects in the Lund configuration spent 6.5% of the time in “red”. A mean difference of 4.7% in “red” signal quality status was found, with the Mason-Likar configuration having less “red” signal time. Using a 95% confidence interval, this difference could be as small as 1.3% but as much as 8.1% of time spent in “red” signal quality (p=0.007).

The Impact of ICU or PCU Status on Signal Quality

A matched-pair t-test was used to determine differences in signal quality between the Mason-Likar and Lund electrode configurations between the ICU and PCU. First, the data were analyzed to evaluate whether there are differences in signal quality between the two electrode configurations based on hospital unit type, ICU or PCU. Analysis then proceeded to compare signal quality between the Mason-Likar and Lund electrode configurations in the ICU and PCU.

Signal Quality in the ICU for Mason-Likar and Lund Electrode Configurations

In the ICU, the mean time spent in a clean "green" signal is not different between the Mason-Likar and Lund electrode configurations as shown in **Table-4.5** (p=0.085). Using a 95% confidence interval, there is a mean difference of "yellow" signal quality time of 3.0%. This difference may be as little as 0.3% but could be as much as 5.7% (p=0.031). There is a statistically significant but relatively small difference of 3% time spent in a "yellow" signal state over the mean 23.8-hour monitoring period. The mean percentage of time spent in a "red" signal status showed an insignificant difference of 0.6% (p=0.747). In the ICU, there is no difference in signal quality between the Mason-Likar and Lund electrode configurations with regard to the percentage of time spent in "green" or "red" signal status.

Table-4.5—Comparison of Signal Quality in ICU Subjects (N=19) Undergoing Continuous ECG monitoring in Both the Mason-Likar and Lund Electrode Configurations

Signal Quality	Mean % Time Mason-Likar (SD)	Mean % Time Lund (SD)	Difference in Mean %Time	p-value*
Mean % Time with Green Signal	71.1 (15.6)	67.6 (16.1)	3.6	0.085
Mean % Time with Yellow Signal	25.1 (11.0)	28.1 (12.7)	-3.0	0.031
Mean % Time with Red Signal	3.8 (15.4)	4.4 (12.7)	-0.6	0.747

* p-value by Match-Paired T-testing

Signal Quality in the PCU for Mason-Likar and Lund Electrode Configurations

Signal quality was compared in the same manner for PCU as for it was for ICU, evaluating the mean percentage of time each patient spent in "green", "yellow", and "red" signal quality in both the Mason-Likar and Lund electrode configurations as shown in **Table-4.6**. In the 81 subjects studied in PCU, patients in the Mason-Likar configuration spent a mean of 9.3%

more time in a “green” signal state compared to the Lund configuration (p<0.001). Using a 95% confidence interval, this difference in mean percentage of “green” signal status time is at least 5.5% but could be as much as 13.2% more time the Mason-Likar configuration spent in “green” signal state than the Lund configuration.

In the PCU, there is a significant difference in “red” signal state between the two electrode configurations, with the Mason-Likar having a mean of 5.5% less time spent with a “red” signal quality state compared to the Lund configuration (p=0.007). Using a 95% confidence interval, this difference in mean “red” signal state is at least 1.3% but may be as much as 8.1%.

Table-4.6—Comparison of Signal Quality in PCU Subjects (N=81) Undergoing Continuous ECG monitoring in Both the Mason-Likar and Lund Electrode Configurations

Signal Quality	Mean % Time Mason-Likar (SD)	Mean % Time Lund (SD)	Difference in Mean % Time	p-value*
Mean % Time with Green Signal	65.6 (17.0)	56.3 (20.1)	9.2	<0.001
Mean % Time with Yellow Signal	33.1 (16.2)	36.8 (17.9)	-3.7	0.029
Mean % Time with Red Signal	1.4 (4.9)	6.9 (18.7)	-5.5	0.007

*p-value by Match-Paired T-testing

Signal Quality Differences for the Lund Configuration between ICU and PCU

The next analyses examine the signal quality of the electrode configurations between the ICU and PCU patient care units. In **Table-4.7**, a comparison of the Lund configuration in both the ICU and PCU shows that there is a mean difference of 11.7% time spent in a “green” signal state between the ICU and PCU, favoring ICU having a better mean “green” signal state (p=0.016). A 95% confidence interval for this difference of mean “green” signal quality time may be as small as 2.4%, but may be as much as 21.0%. In the Lund electrode configuration, there

is a cleaner signal in the ICU when compared to the PCU in the same subjects. There is no significant difference in the Lund configuration "red" signal state between the ICU and PCU (p=0.530). In the Lund configuration, the ICU had less "yellow" signal quality time than PCU (p=0.015).

Table-4.7—Comparison of Signal Quality for the Lund Electrode Configuration Between the ICU and PCU

Signal Quality	ICU (SD)	PCU (SD)	Difference in Mean % Time	p-value*
Mean % Time with Green Signal	68.0 (16.8)	56.3 (20.1)	11.7	0.016
Mean % Time with Yellow Signal	27.4 (12.9)	36.8 (17.9)	-9.4	0.015
Mean % Time with Red Signal	4.6 (13.2)	6.9 (18.7)	-2.3	0.530

*p-value by Paired t-test

Signal Quality Differences for the Mason-Likar Configuration between ICU and PCU

In the Mason-Likar electrode configuration, there are no differences in "green" or "red" signal states between the ICU and PCU clinical areas as shown in **Table-4.8**. There is a difference between the ICU and PCU with the mean percentage of time spent with a "yellow" signal status, similar in magnitude to the difference in "yellow" signal quality state in the Lund configuration. Although there is a significant difference in "red" signal quality state between the electrode configurations in PCU, this difference does not exist in the ICU. Furthermore, in neither electrode configuration was there a significant difference in the mean "red" signal quality state between ICU and PCU.

Table-4.8—Comparison of Signal Quality for the Mason-Likar Electrode Configuration between the ICU and PCU

Signal Quality	ICU (SD)	PCU (SD)	Difference in Mean % Time	p-value*
Mean % Time with Green Signal	71.1 (16.1)	65.6 (16.9)	5.5	0.194
Mean % Time with Yellow Signal	25.1 (11.3)	33.0 (16.1)	-7.9	0.016
Mean % Time with Red Signal	3.8 (15.8)	1.4 (4.8)	2.4	0.518

*p-value by Matched-Paired t-test

Summary of Unit Differences in Signal Quality

This analysis supports the suppositions that ambulatory PCU patients have a noisier ECG signal than do the higher acuity ICU patients, and that the Mason-Likar configuration is more noise-immune than the Lund. In the ICU, there was really no difference in signal quality between the two proposed electrode configurations. In the PCU; however, there is a significant difference in both “green” and “red” signal quality states between the Mason-Likar and Lund electrode configurations, favoring the Mason-Likar as less noise in the signal. For the Lund configuration, there is a greater mean “green” signal state in the ICU, but Mason-Likar showed no difference in “green” signal state between the unit types.

False Lethal Arrhythmia Alarms by Electrode Configuration

Aim 2: ***In patients in the ICU or PCU, determine whether the Mason-Likar and Lund electrode configurations differ in the number of total alarms and false life threatening alarms generated by the monitor system's arrhythmia analysis algorithm over a 24-hour monitoring period.***

Hypothesis 2: There is no difference in false lethal arrhythmia alarms between the Mason-Likar and Lund electrode configurations.

Both the Mason-Likar and Lund electrode configurations will be discussed in this section in the context of false lethal arrhythmia alarms. Lethal arrhythmia alarms included asystole, bradycardia, ventricular tachycardia, and ventricular fibrillation. The arrhythmia alarms data is not normally distributed and can undergo analysis by nonparametric testing only.

Lethal arrhythmia alarms were considered to be asystole (no R-wave for > 6 seconds), ventricular fibrillation, ventricular tachycardia (QRS width > 0.12 sec and ventricular rate >100 beats per minute), and bradycardia (ventricular rate under 50 beats per minute for greater than 6 seconds). There were 2481 total lethal arrhythmia alarms. For ventricular fibrillation, there were only 3 alarms in 2 subjects that were asynchronous during high signal noise states, so the remainder of this analysis will not include false ventricular fibrillation alarms.

True Lethal Arrhythmia Alarms

Of the 2481 lethal arrhythmia alarms in the dataset, there were 25 true alarms. There were 15 true ventricular tachycardia alarms (7 in Mason-Likar, 8 in Lund), and all of these were self-terminating under 15 total beats. Each of the true alarms appeared in both the Mason-Likar and Lund electrode configurations except for one episode that only was detected only in the Lund configuration. This single true ventricular tachycardia alarm occurring only in the Lund

dataset is from a subject that had several other true ventricular tachycardia alarms in both configurations, and there was a noisy Mason-Likar signal at that time. Review of the time-stamped rhythm strips using the Mortara H-Scribe® Holter monitoring software showed that this non-concordant episode of ventricular tachycardia was a single 6-beat run of ventricular tachycardia at 120-130 beats per minute with a QRS width of 0.13 seconds and visible atrioventricular dissociation. There appears to be similar sensitivity between Mason-Likar and Lund electrode configurations for ventricular tachycardia; however, this sample did not yield enough differences in true ventricular tachycardia alarms to permit proper statistical analysis.

There were 10 true bradycardia alarms each case-matched with 5 in Mason-Likar, 5 in Lund, where the heart rate dropped below 50 beats per minute for at least 6 seconds. With the exception of the single true ventricular tachycardia alarm, concordance between Mason-Likar and Lund electrode configurations for true ventricular tachycardia, asystole, and bradycardia alarms appears similar.

Distribution of False Lethal Arrhythmia Alarms

Initially, there were 2481 total true and false lethal arrhythmia alarms in both electrode configurations, and after careful reanalysis, many of these represented at least one monitoring lead off situations. Review of ECG waveforms in Mortara HScribe® data to remove any leads off conditions resulted in a reduction in the total number of alarms to 978. There were a total of 978 false alarms of asystole, bradycardia, and ventricular tachycardia in 53 subjects, with the remaining 47 producing none of this type of alarms. In the overall sample of 100 subjects, the mean values for each false lethal arrhythmia alarm type were driven by a small number of subjects making comparison of means meaningless. In the sample, the median number of alarms is zero for all three types of lethal arrhythmia alarms. Because of the highly skewed alarm frequency data, description of this data can be accomplished only by nonparametric analysis. Examining overall monitoring alarms is relevant clinically, since this is the number of

alarms that a monitor watcher will have exposure during their work shift. Distribution of these alarms by electrode configuration is shown in **Table-4.9**.

Table-4.9—Distribution of False Alarms by Electrode Configuration

False Alarm	Configuration	Total Alarms	Number of Subjects	Total
Asystole	Lund	27	10	62
	Mason-Likar	35	3	
Bradycardia	Lund	304	17	604
	Mason-Likar	300	16	
Ventricular Tachycardia	Lund	159	15	317
	Mason-Likar	158	23	

No Difference in the Overall Number of Arrhythmia Alarms between the Mason-Likar and Lund Electrode Configurations

To determine whether a difference exists between the Mason-Likar and Lund electrode configurations with regard to the total number of false lethal arrhythmia alarms, a McNemar test was performed. To perform this test, the number of alarms occurring in each electrode configuration independently as well as when alarms occurred in time-matched events was determined. For all of the lethal arrhythmia alarms of asystole, bradycardia, and ventricular tachycardia, there were 213 alarms in the Mason-Likar configuration only, 215 alarms in the Lund configuration only, and 550 time-synchronized alarms that occurred simultaneously in both electrode configurations. In the sample of 100 subjects, the McNemar test to detect an overall difference in the total number of alarms between the Mason-Likar and Lund electrode configurations showed no difference ($p=0.400$).

False Asystole Alarms not Different between Mason-Likar and Lund Electrode Configurations

This study examined the Mason-Likar versus the Lund electrode configurations in false alarm rates. In the overall sample of 100 subjects, there is no significant difference in false asystole alarms between the Mason-Likar and Lund electrode configurations by Related-Samples Wilcoxon Signed Ranks Test (p=0.704). The differences in absolute number of false asystole alarms and the mean number of alarms per patient appear large between electrode configurations, but the wide standard deviations and skewed distributions obliterated the differences statistically.

Further analysis of only the subjects that produced asystole alarms was conducted. All 62 asystole alarms were asynchronous, meaning that there was no pattern to the asystole alarms by electrode configurations. To perform a McNemar test to detect a difference in asystole alarms between the electrode configurations, there were 35 asystole alarms in the Mason-Likar configuration, and 28 asystole alarms in the Lund configuration, and no time synchronized alarms occurring in both configurations. The McNemar test showed no difference in the number of asystole alarms between the Mason-Likar and Lund electrode configurations (p=0.450).

False Bradycardia Alarms not Different between Mason-Likar and Lund Electrode Configurations

There were 604 false bradycardia alarms in 21 unique subjects in both electrode configurations. **Figure 4.9** shows that in the Lund configuration, there were 304 bradycardia alarms triggered by 17 subjects, and in the Mason-Likar configuration there were 300 bradycardia alarms triggered by 16 subjects. To perform the McNemar test to detect whether a difference exists between the electrode configurations with regard to false bradycardia alarms, the number of bradycardia alarms in only the Mason-Likar configuration is 131, the number of

bradycardia alarms in only the Lund configuration is 135, and the number of time-synchronized bradycardia alarms occurring simultaneously in both configurations is 338. The McNemar test showed no difference in the number of false bradycardia alarms between the Mason-Likar and Lund electrode configurations (p=0.854).

False Ventricular Tachycardia Alarms not Different Between Mason-Likar and Lund Electrode Configurations

There were a total of 317 false ventricular tachycardia (VT) alarms in 23 unique subjects between the Mason-Likar and Lund electrode configurations combined. There were 159 false ventricular tachycardia alarms in 10 subjects in the Lund configuration, and 158 VT alarms in 20 subjects in the Mason-Likar configuration. There were 212 synchronous false VT alarms that occurring in both electrode configurations. Using these data, the McNemar test was performed to detect a difference between the Mason-Likar and Lund electrode configurations for false VT alarms. There is no difference between the Mason-Likar and Lund electrode configurations with regard to false VT alarms (p=1.00).

Lethal Arrhythmia Alarms and Signal Quality

Many clinicians believe that false arrhythmia alarms are caused by interference in the ECG signal. This analysis only includes lethal alarms without including nuisance type alarms, such as premature ventricular contraction alarms. Since signal quality was measured as described in **Chapter 3-Methods**, and the results demonstrated a difference in signal quality between the electrode configurations, further analysis was performed to determine whether any difference exists in the number and types of false asystole, bradycardia, and ventricular tachycardia alarms related to signal quality measures.

To accomplish this analysis, each of the adjudicated alarms was time-matched to a signal quality measure of green, yellow, or red at the onset of the alarm state. For each type of alarm, a normalized count for each signal quality was computed by counting the number of each

type of alarm in each signal quality state and dividing by the total duration of each type of signal quality per subject. Due to the non-normal distribution of this data, t-tests were not appropriate to compare the mean numbers of alarms; however, a Paired-Sample Wilcoxon Signed Rank test could be used to detect whether differences exist.

There is No Overall Difference between the Mason-Likar and Lund Electrode Configurations in Green or Yellow Signal Quality States

Comparison of the normalized number of asystole, bradycardia, and VT alarms between the electrode configurations was performed. When in the green signal quality state, there is no difference in the overall number of lethal arrhythmia alarms between the Mason-Likar and Lund electrode configurations ($p=0.219$). When the Mason-Likar and Lund configurations were in a yellow signal quality state, there was also no difference in the normalized number of false asystole, bradycardia, or VT alarms ($p=0.325$).

The next analysis was performed to detect whether there is a difference in the overall normalized number of alarms within each configuration across signal quality. For the Mason-Likar configuration, there is no difference in the normalized number of alarms between the green signal quality state and the yellow signal quality state ($p=0.570$). For the Lund configuration, there is also no difference in the normalized number of overall alarms triggered for asystole, bradycardia, and VT dependent upon signal quality state ($p=0.693$).

False Asystole Alarms Are Not Different between Electrode Configuration in Green or Yellow Signal Quality

For the subjects that experienced false asystole alarms in the Mason-Likar, Lund, or both electrode configurations, the alarms were classified by green or yellow signal quality at the onset of the alarm for comparison by electrode configuration. When asystole alarms occur in a green signal quality state, there is no difference in false asystole alarms between the Mason-Likar and Lund electrode configurations ($p=0.250$). When in a yellow signal quality state, there

is also no difference between the Mason-Likar and Lund electrode configurations in the number asystole alarms (p=0.844).

Since a difference exists between the electrode configurations in signal quality, the next analysis examines differences in the normalized number of asystole alarms within each electrode configuration, but across different signal quality. Comparison of the normalized number of asystole alarms when the Mason-Likar is in green compared to the normalized number of asystole alarms when the Mason-Likar configuration is in yellow signal quality shows that there is no difference (p=0.875). For the Lund configuration, the normalized number of asystole alarms in a green signal quality is not different than the normalized number of asystole alarms when the Lund is in a yellow signal quality state (p=0.125)

False Bradycardia Alarms Are Not Different between Electrode Configuration in Green or Yellow Signal Quality

For the subjects that experienced false bradycardia alarms in the Mason-Likar, Lund, or both electrode configurations, the bradycardia alarms were classified by green or yellow signal quality at the onset of the alarm for comparison by electrode configuration. When bradycardia alarms occur in a green signal quality state, there is no difference between the Mason-Likar and Lund electrode configurations (p=0.685). When in a yellow signal quality state, there is no difference between the Mason-Likar and Lund electrode configurations in the number of false bradycardia alarms (p=0.204).

Since a difference exists between the electrode configurations in signal quality, the next analysis examines differences in the normalized number of bradycardia alarms within the electrode configurations, but across different signal quality states. Comparison of the normalized number of false bradycardia alarms when the Mason-Likar configuration is in green compared to the normalized number of bradycardia alarms when the Mason-Likar configuration is in yellow signal quality shows that there is no difference (p=0.733). For the Lund

configuration, the normalized number of bradycardia alarms in a green signal quality is not different than the normalized number of bradycardia alarms when the Lund is in a yellow signal quality state (p=0.068).

False Ventricular Tachycardia Alarms Are Not Different between Electrode Configuration in Green or Yellow Signal Quality

For the subjects that experienced false ventricular tachycardia (VT) alarms in the Mason-Likar, Lund, or both electrode configurations, the alarms were classified by green or yellow signal quality at the onset of the alarm condition for comparison by electrode configuration. When VT alarms occur in a green signal quality state, there is no difference between the Mason-Likar and Lund electrode configurations (p=0.119). When VT alarms occur in a yellow signal quality state, there is also no difference between the Mason-Likar and Lund electrode configurations in the number asystole alarms (p=0.184).

Since difference exist between the electrode configurations in signal quality, the next analysis examines differences in the normalized number of VT alarms within the electrode configurations, but across different signal quality states. Comparison of the normalized number of VT alarms when the Mason-Likar is in green compared to the normalized number of VT alarms when the Mason-Likar configuration is in yellow signal quality shows that there is no difference (p=0.053). For the Lund configuration, the normalized number of asystole alarms in a green signal quality is not different than the normalized number of asystole alarms when the Lund is in a yellow signal quality state (p=0.129). A summary of these results is shown in **Figure-4.10.**

Figure-4.10. Comparison of Normalized Numbers of Asystole, Bradycardia, and VT Alarms Across Electrode Configurations and Signal Quality State

Alarm Type	Electrode Configuration	Normalized Number of Alarms in Green	Normalized Number of Alarms in Yellow	p-value
Asystole	Lund	0.017	0.059	0.125
	ML	0.038	0.025	0.875
	p-value	0.250	0.844	----------
Bradycardia	Lund	0.350	0.143	0.068
	ML	0.284	0.342	0.733
	p-value	0.685	0.204	----------
Ventricular Tachycardia	Lund	0.120	0.278	0.129
	ML	0.122	0.270	0.053
	p-value	0.119	0.184	----------

* p-values derived from Paired-Sample Wilcoxon Signed Test.

Impact of Abrasive Skin Preparation on Signal Quality and False Lethal ECG Monitoring Alarms

Aim 3: ***In patients in the ICU or PCU, determine whether an abrasive skin preparation has an impact on ECG signal quality or false lethal ECG arrhythmia alarms in either the Mason-Likar or Lund electrode configurations over a 24-hour monitoring period.***

Hypothesis 3: There is no difference in signal quality and false lethal arrhythmia alarms with or without abrasive skin preparation in either the Mason-Likar or Lund electrode configurations.

Skin Preparation and Signal Quality

The impact of an abrasive skin preparation in the context of ECG signal quality and lethal arrhythmia alarms will be discussed in this section. In this study, 45 patients were randomized to an abrasive skin preparation, and the remaining 55 subjects had electrodes applied in the usual fashion without abrasive skin preparation. If the patient was randomized to a skin preparation, both the Mason-Likar and Lund electrode configurations received the preparation in that patient.

No Difference in Mean Green Signal Time with Abrasive Skin Preparation

As previously discussed, the Mason-Likar electrode configuration provides a larger proportion of time with a "green" signal quality when compared to the Lund without regard to skin preparation. Although both the Mason-Likar and Lund configurations improved with the intervention of an abrasive skin preparation, there was not a significant improvement in either electrode configuration. Both **Table-4.11** and **Figure-4.1** show the effect of an abrasive skin preparation on both electrode configurations. There was a greater improvement in the Lund configuration, and there was 7.1% more "green" signal time in the Lund group receiving an abrasive skin preparation; however, this difference failed to reach statistical significance

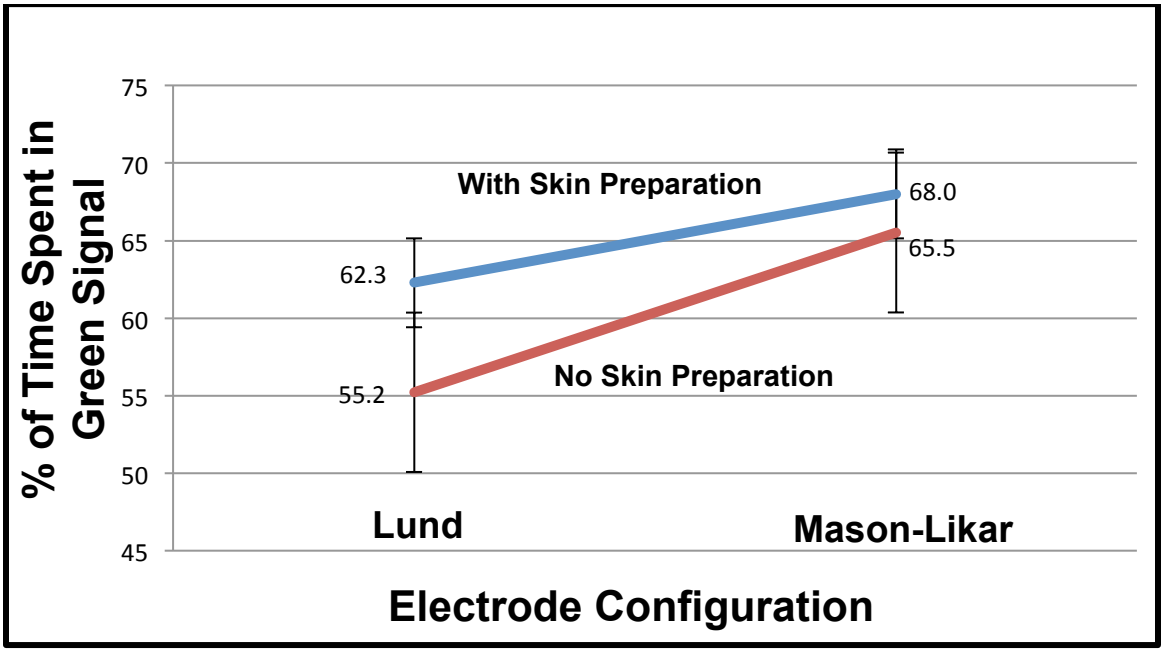
(p=0.077). Similarly, the improvement in “green” signal quality state in the Mason-Likar configuration was smaller than the Lund improvement, with 2.5% more “green” signal quality time with preparation in the Mason-Likar configuration, but this also failed to reach statistical significance (p=0.464).

Both electrode configurations improved the amount of “green” signal quality time when used with a wet-prep electrode. This study did not address signal quality changes that may occur with gel-based electrodes.

Table-4.11—“Green” Signal Quality using an Abrasive Skin Preparation (n=45) by Electrode Configuration

Configuration	Skin Preparation	N	Mean % Green Segments	SD	Difference of Means	p-value*
Lund	YES	45	62.3	19.1	7.1	0.077
	NO	55	55.2	20.4		
Mason-Likar	YES	45	68.0	17.1	2.5	0.464
	NO	55	65.5	16.7		

Figure-4.1--Mean Percent Time with "Green" ECG Signal Quality by Electrode Configuration both with and Without Abrasive Skin Preparation



No Reduction in "Red" Signal Quality Time with Abrasive Skin Preparation

Although there was not a demonstrated significant benefit to increasing "green" signal quality time, the converse question was raised whether there is a difference in the mean amount of time spent in "red" signal quality in each electrode configuration. Abrasive skin preparation did not produce a significant difference in the amount of time spent in a "red" signal quality in either electrode configuration as shown in **Table-4.12.** Again, there was a larger improvement for the Lund configuration in the mean amount of "red" signal quality time with the use of an abrasive skin preparation, although this did not reach statistical significance. The Mason-Likar showed a insignificant 1.06% increase in the "red" signal quality time when an abrasive skin preparation was used.

Table-4.12—Impact of Abrasive Skin Prep and Electrode Configuration on Mean Percent of "Red" Signal Quality Time

Configuration	Skin Preparation	N	Mean % Red Segments	SD	Difference of Means	p-value*
Lund	YES	45	3.76	13.1	-5.02	0.161
	NO	55	8.78	20.7		
Mason-Likar	YES	45	2.41	10.4	1.06	0.515
	NO	55	1.35	5.5		

*p-value by t-test

Abrasive Skin Preparation and False Lethal Arrhythmia Alarms

Skin Preparation and False Lethal Arrhythmia Alarms

There is a supposition among clinicians that false alarms are related to poor signal quality, and skin preparation is aimed at improving signal quality. The second part of Specific Aim 3 is to determine whether an abrasive skin preparation changes the number of false lethal arrhythmia alarms.

By patients wearing both electrode configurations, these groups are related samples, and these analyses were performed using a Mann-Whitney test. Because the false lethal arrhythmia alarms in the Lund configuration are independent of the false lethal arrhythmia alarms in the Mason-Likar configuration, a more conservative option for treating the alarms data is the Related-Samples Friedman's Two-Way Analysis of Variance by Ranks. Using the

Friedman's statistic, in neither the Lund nor the Mason-Likar were any of the false lethal arrhythmia alarms significant.

False Asystole Alarms not Different with Abrasive Skin Preparation

Due to the non-normal distribution of the highly skewed alarms frequencies, data were analyzed by nonparametric techniques. The Mann-Whitney tests were performed within each configuration across abrasive skin preparation status. The Lund electrode configuration has no difference in false asystole alarms between patients that underwent abrasive skin prep and those who did not ($p=0.700$). The Mason-Likar electrode configuration also has no difference in false asystole alarms between patients that underwent abrasive skin prep and those who did not ($p=0.702$). . Comparison of asystole alarms between electrode configurations showed that only 6 of the 62 total asystole alarms occurred in subjects that underwent abrasive skin preparation but a nonparametric Mann-Whitney testing did not show any significant results between skin preparation status and false asystole alarms that occurred in any state of signal quality.

False Bradycardia Alarms and Abrasive Skin Preparation

There were 10 true bradycardia alarms where the heart rate dipped below 50 beats per minute for at least 6 seconds, and there were 604 false bradycardia alarms, with 304 Lund false bradycardia alarms in 17 patients, and 300 false bradycardia alarms in 16 subjects. Skin preparation was examined as a potential method to reduce false bradycardia alarms. In the Lund electrode configuration, there is no difference in false bradycardia alarms between patients that underwent abrasive skin prep and those who did not ($p=0.219$). In the Mason-Likar electrode configuration, there is also no difference in false bradycardia alarms between patients that underwent abrasive skin prep and those who did not; however, this result approaches statistical significance ($p=0.058$). When comparing the effect of skin preparation between the Mason-Likar and Lund electrode configurations, the Mann-Whitney test was performed that

showed no significant differences in false bradycardia alarms between electrode configurations across skin preparation status.

False Ventricular Tachycardia Alarms and Abrasive Skin Preparation

In both the Mason-Likar and Lund electrode configurations, there were 317 false ventricular tachycardia and 15 true ventricular tachycardia alarms. The Lund had 159 false ventricular tachycardia alarms in 15 subjects, while the Mason-Likar electrode configuration had 158 false ventricular tachycardia alarms in 23 subjects. Skin preparation was examined as a potential method to reduce false ventricular tachycardia alarms.

The Lund electrode configuration shows no difference in false ventricular tachycardia alarms between patients that underwent abrasive skin prep and those who did not ($p=0.634$). In the Mason-Likar electrode configuration, there was no difference in false ventricular tachycardia alarms between patients that underwent abrasive skin prep and those who did not ($p=0.668$). Comparison of the number of false ventricular tachycardia alarms between the Mason-Likar and Lund electrode configurations by preparation by nonparametric Mann-Whitney testing showed no significant differences.

Patient Perceptions, Attitudes, and Suggestions for Continuous In-Hospital ECG monitoring

Aim 4: *Describe patient perceptions of each electrode site with regard to comfort, interruption to sleep, and impairment of movement, as well as general perceptions of the experience of being monitored.*

The information gathered from patients at enrollment, during the monitoring period, and at the removal of the study devices was initially unsolicited; hence, less than 90 subjects were able to provide this qualitative data that will be presented in a mixed qualitative-quantitative approach. In addition to the original qualitative questions regarding whether any electrode sites

were bothersome, themes developed on the topics of monitoring as given by unsolicited patient remarks. The following domains of analysis were developed through interviews. These domains included the electrode placement locations, monitoring equipment, and the perception of being continuously monitored. Due to the nature of the data, only descriptive statistics can be applied.

Arm Placed Electrodes Do Not Bother Most Patients

Part of this study included analysis of patient factors to determine the feasibility of using the Lund electrode configuration for continuous in-hospital ECG monitoring. In this study, all subjects wore two Holter monitors that were attached in the Lund and Mason-Likar electrode configuration. At the end of data collection, subjects were asked to point to any electrodes in either configuration that were particularly bothersome. Of 94 subjects offering an opinion, only four (4.26%) felt that the right and left arm electrodes of the Lund configuration were bothersome at all, and the subjects reported only mild bother. No subjects reported that the electrodes in the Mason-Likar configuration were bothersome. Bothersome was defined as limiting activity, sleeping, or performing activities of daily living. In achieving this study's overarching goal of determining the ideal electrode configuration for continuous in-hospital ECG monitoring, placing the electrodes on the outer arm at or below the level of the axillary fold does not represent a limitation from a patient comfort perspective.

Patients Feel "Safer" But More Confined While Being Monitored

In this study, there were three subjects who stated that they felt safer because they knew that the hospital staff was "watching their heart". There were two ICU subjects that reported that although the arm electrodes did not bother them in either configuration, the monitor beeping woke them up so frequently at night that they could not sleep. One patient described being on a telemetry monitor like being on a leash, since he could not leave the cardiac progressive care unit. Three subjects requested that the monitoring equipment be

waterproof so that they could shower with it still attached. All of these patients had been disallowed from showering so that they could remain continuously monitored.

During the study enrollment period, there were three unique subjects that experienced panic attacks so severe as to require removal of the study monitors and hospital monitoring equipment. One of these three signed out of the hospital against medical advice, reporting to study staff that he wanted to be able to smoke cigarettes. One panic attack subject remained hospitalized but unmonitored. The third panic attack study participant approached study staff during their next data collection day and asked to participate in the study since she was feeling "less anxious". This patient subsequently exceeded the study-monitoring period by providing 26 hours of continuous data. A common trait observed in all three panic attack subjects was active tobacco use at the time of admission to the hospital and enrollment in the study.

Patient Perceptions of Continuous ECG Monitoring Equipment

Hospital ECG equipment has shrunk is size over the past few decades; however, patients still view being monitored as a burdensome component of the hospital experience. For the purposes of this study, patient responses regarding hospital monitoring equipment was broken down into three component parts: (Sanford P. Bordeau (1982) Volts to Hertz...the Rise of Electricity. Burgess Publishing Company) the monitoring connection in the ICU or the telemetry box for the PCU patients, (Sanford P. Bordeau (1982) Volts to Hertz...the Rise of Electricity. Burgess Publishing Company) the electrode wires and electrode attachment, and (3) the skin electrodes.

Patients Feel Telemetry Boxes are Too Heavy To Wear

In the ICU, the patient is typically "hard-wired" to their bedside monitoring equipment. Of the 19 ICU subjects in this study, all reported that the heavy weight of the cable coupling the electrode wires to the wiring harness of the bedside monitor was too heavy. Subjects believe that the wires do not need to be this heavy to monitor their heart. All subjects in the ICU

reported that the electrode wires were “too stiff” and frequently popped out of the gown into their face, and four subjects said the wires were so stiff that the wires pushed the electrodes off of their skin.

In the PCU, patients made numerous comments regarding the telemetry box. In this study, 85 subjects (four ICU patients transferred to PCU during their study interval) reported on their perceptions of the telemetry box. The boxes were reported as “too heavy” by 57.0% of subjects, with 32.9% commenting that the boxes were too large. Only 7.14% remarked that the shape of the box is an issue. The patients request that subsequent monitor designs are lighter in weight and smaller in size. Having also seen the size, shape, and felt the weight of the Holter monitors used in this study (Mortara H12+, Mortara Instruments), subjects felt that the size, shape and weight of these devices was more conducive to a comfortable monitoring experience.

Methods to Carry the Telemetry Boxes

Numerous unsolicited comments were made by participants regarding the method for carrying telemetry boxes, with 39 of 81 patients making comments and suggestions to reduce the cumbersome nature of carrying a telemetry box. Participants in the study were given a woven paper neck pouch to carry study Holter monitors, while the standard method by the hospital is to use a pocket in the hospital gown pocket. Comments made by patients were very candid such as shown in **Table-4.13**.

Table-4.13—Comments and Suggestions from Patients To Improve Carrying the Telemetry Box

Number of Subjects Reporting	Comment
12	This neck pouch is better than the hospital gown pocket.
8	Why can't we have pants? I'd like to just put this in my pocket.
5	I hate this neck strap—can you make it softer?
4	This box keeps hitting my incision in the middle of my chest.
3	I would like to wear pants, but my incision is on my waistline.
3	This pendant pouch is always swinging around getting in the way.
2	This box in the gown pocket is rubbing my nipple in not a good way.
2	My central line is getting irritated with this neck strap.

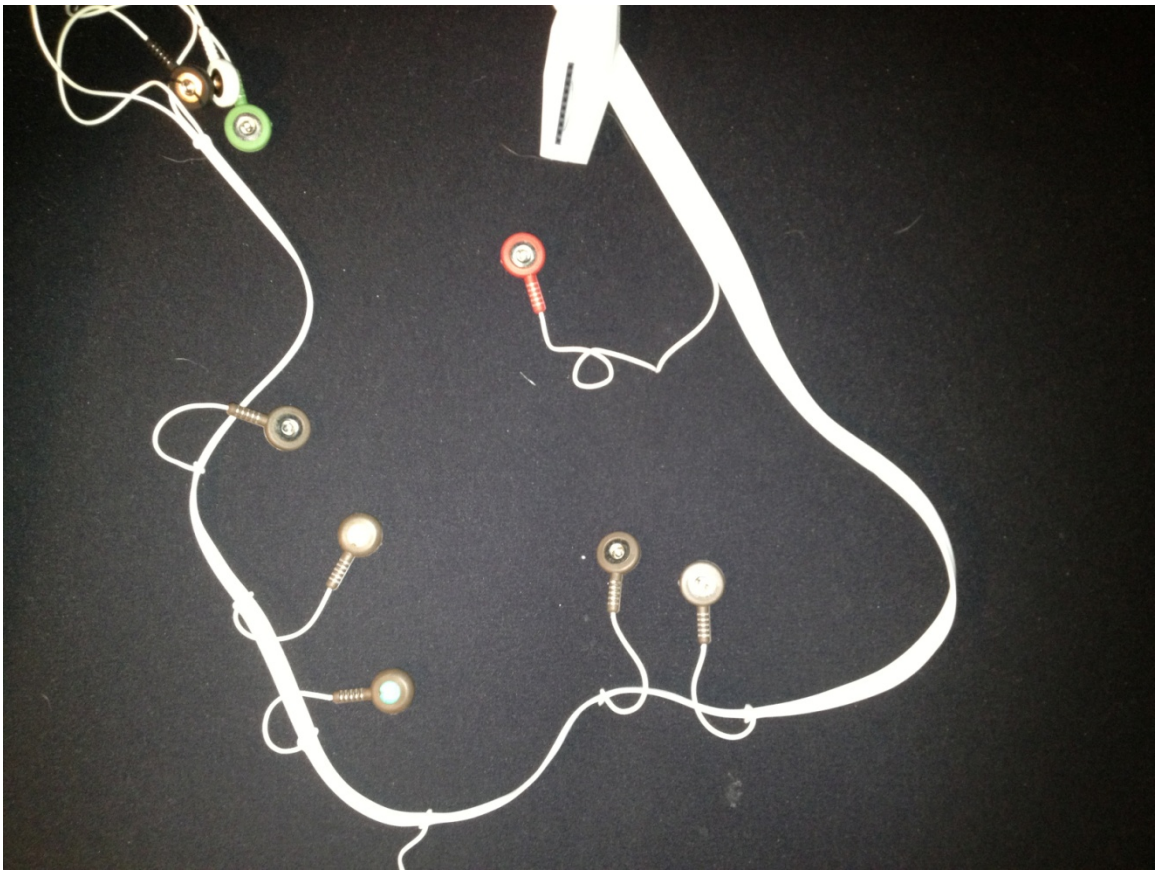
There were two subjects that raised a very valid question. They both stated that they were never offered options to carry the boxes. In our sample, only 8 patients were wearing pants, and many male subjects would prefer to wear the telemetry box in a pocket or a belt clip. Male subjects were more likely to complain that they did not like the neck strap or a gown pocket strategy to carry the telemetry box.

Electrode Lead Wires Are Bothersome

In this study, 54% of patients made comments about the electrode wires that connect the monitoring device to the skin electrodes. Comments were both positive and negative drawing on their experience from participation in this study. Tangling, stiff wires popping out of the neck of the gown, and pain while lying or sleeping on the electrode wires was reported by 23% of respondents. Interesting descriptors of the electrode wires perception included, "It feels like an octopus on me", "Why cant' these things have no wires at all", and " I wish they would color code these wires and electrodes so that the patient could reconnect them when they come

off". The electrode wires used on the study Holter monitors is a single cable design that allows each electrode wire to be "peeled" off of the trunk wire as shown in **Figure-4.2**. Additional comparative comments were that 28.7% of patients stated that the study electrode wires were more comfortable. Another three subjects expressed a need for longer wires to prevent "pop-offs".

Figure-4.2—Patients preferred this type of single trunk electrode lead wires with branches that "peel" off (Mortara Instruments, Inc.)



Electrodes Impact Patient Comfort

At the end of the data collection intervals, 92% of the 100 subjects had their study monitoring systems removed by study staff, the remainder was removed either by the patient, nursing staff, or monitoring technician due to early discharge. At the end of the study interval, there was a mean of 11.8 intact electrodes of the 12 required for the study. For this study, it is important to differentiate electrode dislodgement from when an electrode wire disconnects from

the electrode. There were 15 known electrode wire disconnects from the electrode and 11 electrode dislodgements from the skin surface. Of the electrode wire disconnects, 11 of the 15 occurred at the Left Leg electrode site with the remaining scattered on the arms and chest. The frequency of electrode dislodgements from the skin is described in **Table-4.14**.

Table-4.14--Electrode Dislodgements by Site for the Mason-Likar and Lund Electrode Configurations

Electrode Site	Mason-Likar Configuration	Lund Configuration
LL	5	3
RL	2	1
V1	1	0
V5	2	1
Total Dislodgements	10	5

Electrode Snaps versus Clips

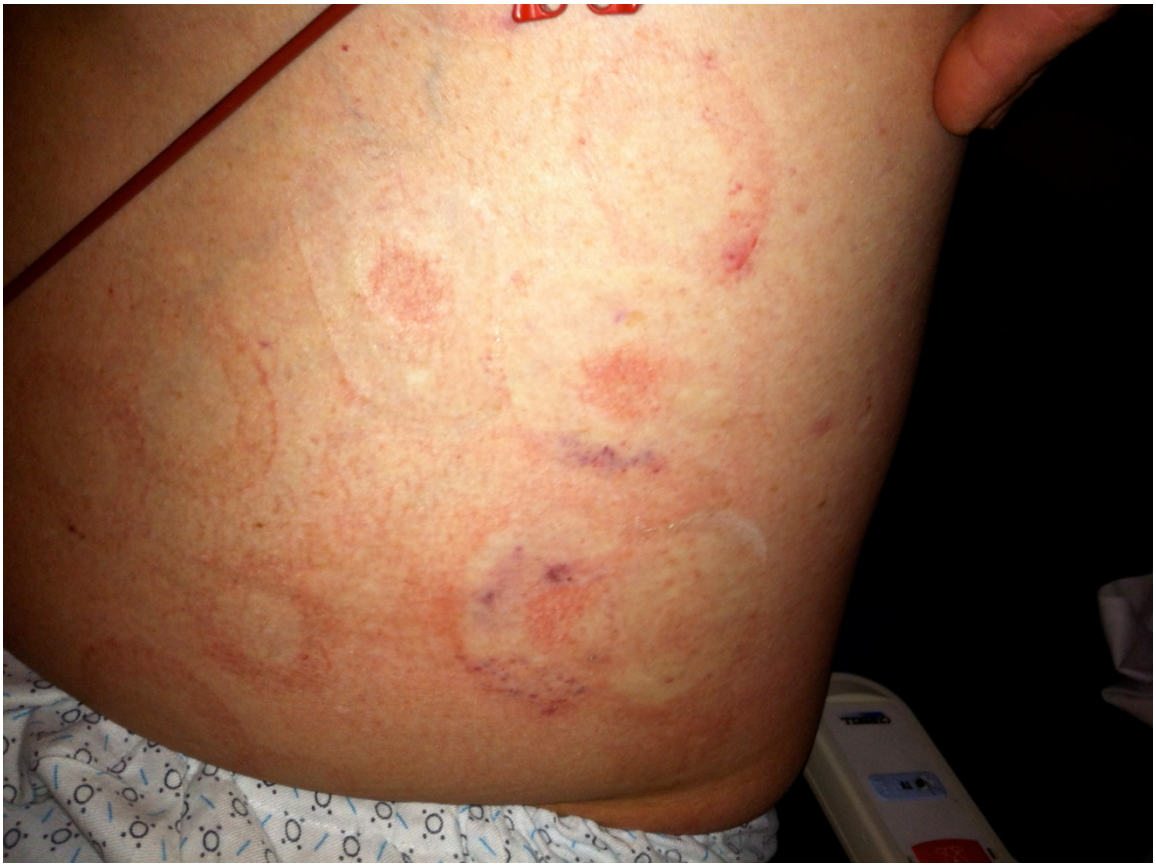
A small proportion of 7 patients expressed comments about the mechanism by which an electrode connects to the wires to the ECG monitoring unit. Patients reported that the hospital-used clips that connect the wire to the electrode disconnect more frequently than the press-on snaps. However, study patients that experienced a snap disconnect from the study monitor electrodes reported discomfort if the snap were pressed into an electrode already present on the skin, and they commented they would prefer if the snap were attached to a new electrode prior to being placed on their skin.

Painful Electrode Removal, Skin Reactions, and Shaving for Electrode Placement

In this study, 71.3% (62 of 87 patients) reported at least moderate discomfort with electrode removal; however, 56.5% (35 of 62) of these subjects remarked that the wet-prep

electrodes induced less discomfort with removal when directly compared to the hospital-used gel-backed electrodes. Painful removal of electrodes was a frequent occurrence. There were two subjects that were approached to participate in the study but declined to participate due to known allergic reaction to the electrodes, although neither could state whether they are allergic to the adhesive or the electrode conductant gel. One of the two subjects agreed to participate in a test since she was hospitalized for an arrhythmia, had prior radiofrequency ablation of an arrhythmia, and she was not being monitored in the hospital due to the electrode allergy. She did react to gel conductant on the hospital's gel-backed electrode, but did not react to the wet-prep conductant or adhesive. Another subject did report itching from all electrode sites, both hospital monitoring and study electrode sites at the end of the study. This gentleman reacted to the adhesive of the wet-prep electrodes and the gel conductant of the hospital electrodes as shown in **Figure-4.3**.

Figure-4.3—Skin Reaction to Adhesive of Study Wet-Prep and Gel Conductant of Hospital Electrodes—various sites show reactivity to the adhesive perimeter of the Wet-Prep electrodes and the conductant gel in the Gel-Backed electrodes.



Some patients, particularly men, had hairy chests and arms that made good adhesion and contact for the electrodes nearly impossible. To participate in the study, shaving was required for any subject that had body hair in electrode locations; however, the hospital did not require this action to be taken for routine monitoring. In this study, only 8 male patients required shaving at the electrodes sites to participate in the study. In our post-participation survey, the male subjects were asked how they feel about being shaven prior to electrode placement. In our study, 71.1% of the sample's male subjects would prefer to be shaven prior to electrode placement to prevent a painful removal. Comments made by three patients were to request that electrodes would be like commercially available "Command® (3M)" strips that release the

adhesive by a proprietary mechanism.

Results Summary

The overarching goal of this study is to determine the ideal electrode configuration for continuous in-hospital cardiac monitoring. The Mason-Likar torso positioned electrode configuration is popular and widely-used for stress testing and cardiac monitoring; however, to improve faithfulness to the standard ECG, the arm-positioned Lund electrode configuration was proposed. If the Lund configuration were to be accepted into practice standards, it would need to be as good as or better than the Mason-Likar configuration with regard to signal quality, monitoring alarms, and patient comfort in a dynamic real-world setting.

Chapter 5. Conclusions and Discussion

Introduction

This is the first study to be able to objectively and reliably quantify similarities and differences of the Mason-Likar and Lund electrode configurations regarding signal quality. Important relationships between signal quality, false lethal arrhythmia alarms, patient comfort and perceptions, skin preparation and the interaction of all these factors were also evaluated. This comparison of the widely popular Mason-Likar to the proposed Lund electrode configuration is vital in evaluating whether the potential adoption of the Lund configuration into continuous monitoring and stress testing is feasible or necessary. The advantage of the Lund configuration is that the ECG waveform more faithfully represents the diagnostic 12-lead ECG than does the Mason-Likar configuration. Prior to this study, it was unknown whether differences exist between these electrode configurations with regard to signal quality, false lethal arrhythmia alarms, and patient comfort. This study is important to propel ECG science forward since the methods are novel applications of widely accepted, FDA-approved technologies in real-time signal quality over a typical 24-hour monitoring interval. This study is unique in using these technologies in continuous ECG monitoring situations, which go beyond the standard 10-second diagnostic ECG.

One of the most important aspects of understanding the impact of this study is related to the acquisition of a diagnostic 12-lead ECG using bedside monitoring equipment. Although monitoring all 12-leads with 10 electrodes is possible, it is typically not performed in all patients due to patient comfort, equipment limitations, and unclear outcomes benefits. Current technology used for continuous monitoring allows for the addition of the six precordial electrodes (V1-V6) to the usual 5 electrodes used for continuous monitoring to acquire a full 12-lead ECG from the bedside monitoring equipment. The problem has little to do with the addition of the precordial electrodes, but rather when the diagnostic ECG is acquired with limb

electrodes that are in non-standard ECG locations, such as the Mason-Likar or Lund configuration. Although included in the directions for use by manufacturers, many staff in clinical practice do not routinely move the arm and leg electrodes from the proximal, centrally located monitoring sites for limb electrodes to the proscribed distal locations as specified for a proper standard, diagnostic ECG (Drew et al., 2004)

For PR-QRS-QT interval assessment, all three approaches are equivalent. For arrhythmia interpretation, there are subtle differences mostly related to the changes in axes that occur with alternate electrode locations, as well as difference in the detection of prior ischemic events through Q-wave identification. The lack of faithfulness to the standard 12-lead ECG from modified placement of limb electrodes can interfere with proper detection of Q-waves and inferior ischemia, presence or absence of ischemia or myocardial damage, and has led to unnecessary risk and procedures for patients (Knight, Pelosi, Michaud, Strickberger, & Morady, 1999, 2001). The Lund electrode configuration was proposed as a potential solution for using bedside monitoring equipment to produce 12-lead ECG's using bedside monitoring equipment that is more faithful to the standard ECG, without the concern over moving limb electrodes to standard locations.

The similarities and differences between the Mason-Likar and Lund electrode configurations in a continuous monitoring situation were not clear prior to this study. In continuous monitoring, it was unclear whether moving limb electrodes onto the arms, as proposed in the Lund electrode configuration, would result in a degraded signal quality due to motion artifact with arm and leg movement, as well as myoelectric artifact with patients lying on the limb electrodes. It was also not known whether there would be a difference in the number and types of false lethal arrhythmia alarms between the two electrode configurations. Skin preparation for electrode placement has been studied in peculiar ways on healthy volunteers that are not directly applicable to continuous ECG monitoring situations. This study incorporated a randomized abrasive skin preparation into the real-world design using actual hospitalized

patients in both simultaneous electrode configurations.

As data collection in this study began, patient perceptions of the monitoring experience was something that the patients wanted to provide. Initially, subjective data regarding the bother of arm-placed electrodes was the initial aim of the post-study interview; however, patients began offering input into improving the next generation of monitoring equipment and techniques. This qualitative data was interesting and valuable for improving the patient experience of being monitored.

Format for Discussion and Conclusions

1. Average ECG signal quality is different based on electrode configuration.
2. Signal Quality Differences between ICU and PCU.
3. ECG signal quality did not impact the numbers or types of false lethal arrhythmia alarms.
4. No difference in false lethal arrhythmia alarms between the two electrode configurations.
5. Abrasive skin preparation made positive but insignificant changes in ECG signal quality.
6. Strengths and weaknesses of this study design, measurement tools, and data analysis.
7. Recommendations for ECG monitoring clinical practice.
8. Recommendations for future research to reduce the alarm fatigue in clinicians.

Average ECG Signal Quality is Different between Electrode Configurations

Prior to this study, it was believed that there was a difference between the two electrode configurations in signal noise. Prior work relied on visual inspection by three raters on a 5-point Likert scale, and these scientists concluded that the Lund ECG tracings were as noise-immune as the Mason-Likar with improved faithfulness and similarity to the standard 12-lead ECG

(Welinder et al., 2004). The aforementioned Welinder, et al (2004) study showed similar visual ECG waveform noise immunity between the Mason-Likar and Lund electrode configurations in 20 healthy subjects providing 6 asynchronous 10-second ECG's each. Our study found that there is a significant difference in ECG signal noise between the Mason-Likar and Lund electrode configurations. There may be methodological or measurement issues that explain the differences between automated signal noise measurement and human-visual estimation of ECG signal quality.

In our study of 100 patients, each patient provided a mean of 23.8 hours of continuous data with synchronous, simultaneous recording of both electrode configurations. A reliable, computerized algorithm that determined the number of "green-yellow-red" segments for the electrode configurations then analyzed these ECG signals. Significant differences were found in the amount of ECG signal quality between the Mason-Likar and Lund electrode configurations favoring the Mason-Likar. The Mason-Likar configuration provided an average of 8.2% for more "green" signal time compared to the Lund electrode configuration ($p<0.001$). Also, there was a difference favoring the Mason-Likar over the Lund electrode configuration with regard to the amount of time spent with a "red" level of signal quality ($p=0.007$).

This "red" level of signal quality is clinically important since the monitoring algorithm enters an "arrhythmia suspend" status during this time, which deactivates all arrhythmia alarms including lethal arrhythmia alarms. Time spent in a "red" signal quality is essentially unmonitored time. Over the mean monitoring period of 23.8 hours, the Lund configuration spent a mean of 6.5% (92.8 minutes) of the time in a "red" signal quality state, while the Mason-Likar configuration had a mean of 1.8% (25.7 minutes) of their time in "red". Future analysis will be necessary to evaluate more about the characteristics of the duration and frequency of the "red" signal quality periods as part of a determination of risk management for missed arrhythmia alarms.

Unlike the previously cited work by Welinder, et al (2004), this study was also the first to

use an industry-accepted, FDA-approved signal quality algorithm (Hook-Up Advisor®) to objectively and reproducibly quantify the quality of signals without interrater bias. This adaptation of Hook-Up Advisor® represents technological novelty for this study in the aspect of signal quality measurement, not estimation, as in prior work. Use of visual inspection of ECG waveforms for signal quality is subjective; however, there is not a defined "gold standard" to measure signal quality in the context of waveform interpretation by human or computerized analysis. Computerized analysis has typically been limited to the abilities of the human programming the algorithm; however, in the modern era of machine learning, this may come into question. It is unclear from this study whether the human or computerized assessment represents the best interpretation in clinical context. Furthermore, this study provided analysis of data from running the raw ECG signals through the Hook-Up Advisor® only one time, so determining the reliability of the algorithm is limited. One area for future research is to examine the reliability of human and automated ECG signal quality analysis. From this dataset, future research will focus on the differentiation of baseline wander, myoelectric noise, and electrode failure will be valuable for improving signal acquisition strategies.

In the United States, one of the major reasons cited for not using automated ST-segment monitoring and the alarms associated with it relates to the concern over generating high numbers of false ST-segment alarms (Crater et al., 2000; Drew & Funk, 2006). The amplitude and morphology of the ST-segment is subject to a certain amount of variability based on body position, and the ST-segment is very dynamic (Adams & Drew, 2002; Adams, Pelter, Wung, Taylor, & Drew, 1999). On average, the Mason-Likar electrode configuration offers a cleaner ECG signal, with more time spent in a “green” signal when compared to the Lund electrode configuration. In stress testing, it may be more important to have a clean and clear ECG signal for real-time automated and human analysis of ischemia. Two major goals of continuous monitoring are arrhythmia recognition as well as ischemia identification, and improving signal quality may play a role in automated ST-segment analysis for more accurate

and timely ischemia detection. A limitation of this study is that ST-segment alarms were not a specific aim, and future work should be aimed at the impact of signal quality and electrode configuration on human and automated analysis of ST-segments.

The original goal of the Mason-Likar configuration was to find a more noise-immune ECG tracing for ST-analysis in stress testing, and this study finds that the Mason-Likar configuration is successful in providing a better average signal quality in a continuous monitoring situation (Mason & Likar, 1966). A future question is whether the improvement in signal quality and noise immunity outweighs the technical limitations associated with the loss of faithfulness to the standard 12-lead ECG and the known differences in axes and Q-waves in real-time arrhythmia and ischemia analysis (Farrell, Syed, & Gutterman, 2008; Kleiner, Nelson, & Boland, 1978; Madias, 2006; Pahlm & Wagner, 2008; Papouchado, Walker, James, & Clarke, 1987; Welinder et al., 2004). In clinical practice, it may be worthwhile to forgo faithfulness to the standard ECG, and accept altered axes and Q-waves in alternate electrode configurations temporarily to improve the accuracy of detecting ischemic changes in the ST-segment. Using bedside equipment to obtain diagnostic ECG's may be a feasible concept, but the practice presents challenges to have staff correctly relocate limb electrodes to the standard locations from proximal monitoring locations.

Signal Quality Differences between ICU and PCU

Prior to this study, it was unclear and unknown whether the more acute intensive care unit (ICU) patients had similar signal quality characteristics compared to their more ambulatory, but lower acuity, progressive care unit (PCU) patients. Prior researchers suggested that clinician interventions were a source of significant numbers of false monitoring alarms, proposing that monitoring alarms were related to ECG signal noise (Tsien, 1997 #52)(Siebig, 2010 #364)(Lawless, 1994 #30). To determine unit-based differences, this study was conducted on ICU and PCU patients with simultaneously collected ECG waveforms from the Mason-Likar and

Lund electrode configurations.

In the ICU, there is no significant difference between the Mason-Likar and Lund electrode configurations for “green-yellow-red” signal quality states. In the PCU, however, there are significant differences between the electrode configurations in signal quality. In the PCU, there is a mean of 9.2% more “green” signal quality time in the Mason-Likar ($p<0.001$) than in the Lund configuration. Time spent with a poor ECG signal quality in the PCU is a mean of 5.5% more “red” signal quality time in the Lund electrode configuration. These findings support the claim that the Mason-Likar electrode configuration is more noise immune in ambulatory PCU patients than is the Lund; however in the ICU it did not make a difference.

Comparing the ICU and PCU patients in the Lund electrode configuration, there is a mean of 11.7% more “green” signal time in the ICU patients than in the PCU ($p=0.016$). There was no difference in the mean amount of “red” signal time between ICU and PCU in the Lund. For the Lund configuration, there is a difference in “green” time, but not in “red” time between the ICU and PCU, suggesting that the ECG noise introduced by the ambulatory patients is mild to moderate, not severe. For the Mason-Likar electrode configuration, no differences were found between the ICU and PCU for the mean amount of “green” signal time ($p=0.194$) or the mean amount of “red” signal time ($p=0.518$).

To summarize the findings regarding signal quality and electrode configuration, the Mason-Likar electrode configuration offers a significant advantage in ECG signal quality; however, it is also known that the Mason-Likar configuration does not produce ECG waveforms that are faithful to the diagnostic ECG. Although both elements are important, it is unclear which is more valuable in clinical monitoring—signal quality or faithfulness to the standard ECG. The assessment tool for ECG electrode configurations promulgated by Drew (2011) assigns a weight of 6 of the total 20 points available to equivalence to the standard ECG. The role of continuous monitoring is to detect the dynamic nature of rhythms and ischemia, and the most significant differences on the ECG waveforms with the Mason-Likar configuration is an altered

axis and deletion of inferior wall Q-waves. It is unclear whether axis and Q-waves are important to detect continuously and dynamically. Future research should include work toward achieving both goals simultaneously—faithfulness to the standard ECG for comparison and trending, as well as a simple electrode application and configuration strategy that provides a more noise-immune ECG signal.

No Difference in False Lethal Arrhythmia Alarms between Mason-Likar and Lund Electrode Configurations

Before this study, it was unknown whether electrode configuration plays a role in the number and type of lethal arrhythmia alarms that a patient would experience in a typical day of monitoring. There is a long-standing assumption that monitoring alarms are caused by interference in the ECG signal, and this study did find a significant difference in signal quality between the electrode configurations. In our study, it does not appear that false lethal arrhythmia alarms are different between the Mason-Likar and Lund electrode configurations. Furthermore, with the majority of alarms being found in the minority of patients, those patient factors contributing to high alarm rates are yet to be identified. A limitation of this research is that it focused on lethal arrhythmia alarms and did not include nuisance alarms, such as premature ventricular contraction alarms. When determining the ideal electrode configuration for continuous in-hospital ECG monitoring, there was no difference in the false lethal arrhythmia alarms between the Mason-Likar and Lund electrode configurations. Future research should examine patient factors related to the number and types of false arrhythmia alarms.

ECG Signal Quality Did Not Impact Number or Types of False Lethal Arrhythmia Alarms

Prior to this study, it was known that arrhythmia alarms have a very high false positive rate (Aboukhalil, Nielsen, Saeed, Mark, & Clifford, 2008; Biot, Carry, Perdrix, Eberhard, & Baconnier, 2000; Biot, Holzapfel, Becq, Melot, & Baconnier, 2003; Chambrin, 2001; Chambrin et al., 1999; Lawless, 1994; Tsien & Fackler, 1997). This study did not aim to determine a false

positive and true positive alarm rate, but this analysis will be planned for subsequent analysis. False lethal arrhythmia alarms were defined as asystole, bradycardia under heart rate of 50 for at least 6 seconds, ventricular tachycardia, and ventricular fibrillation. There were several patients that experienced deterioration requiring escalation of care to the ICU, two subjects requiring emergency surgery, and one requiring a cardiac catheterization. This study analyzed nearly 24 hours of continuous ECG monitoring data in both Mason-Likar and Lund electrode configurations on most of the 100 subjects. Despite the high acuity, there were no subject deaths while enrolled in the study. There were only three ventricular fibrillation alarms in the entire dataset, so there is too little data to offer meaningful analysis of false ventricular fibrillation alarms. Further research with a larger catchment of patients over a longer monitoring period may be able to capture adequate numbers of arrhythmogenic cardiac arrests for a meaningful analysis.

Measurement of the signal quality data was provided by computerized analysis, and this data was a continuous variable with a normal frequency distribution allowing for parametric statistical analysis. The false lethal arrhythmia alarm data; however, is very skewed, limiting analysis to nonparametric techniques. All of the lethal arrhythmia alarms were produced by less than half of the sample, leaving over half of the sample without any lethal arrhythmia alarms. As more research emerges to examine monitoring alarms with the intent to improve the monitoring alarm crisis and decrease alarm fatigue in clinicians, clear measurement units will need to be developed for direct comparison of research data. Whether data should be reported as alarms over time, alarms per patient over time, alarms per bed over time, or alarms requiring human responses are examples of context-sensitive units of analysis. The concept of monitoring alarm burden or clinical alarm burden needs context to relate the various metrics about alarms to patients, providers, or monitor watchers.

In this study sample, the mean number of false lethal arrhythmia alarms per patient appears low because the alarms data is driven by a small subset of the sample that produced

the vast majority of alarms. Previously cited research suggested that staff manipulation of patients and ambulatory patients created a corrupt ECG signal fraught with noise, and this noise created false arrhythmia alarms (Tsien, 1997 #105(Lawless, 1994 #104)). This study found that in the ICU, where there is presumed to be more patient manipulation by staff, there was less ECG signal noise. This study was conducted in adult acute care areas, where the previously cited studies were conducted in pediatric acute care areas. In this study, no association between worsened signal quality and increased false arrhythmia alarms could be established.

Overall, there was a low number of each type of alarm per patient over the mean monitoring interval of 23.8 hours as shown in **Table-4.9**. Although these numbers seem small, in clinical context, a monitor watcher typically does not watch a single patient. Monitor watchers a typically responsible for a larger pool of patients. Using this dataset as an example of a monitor watcher's alarm burden for 100 subjects in just the Mason-Likar configuration, there would have been 493 alarms in less than 24 hours. In a typical 8-hour shift, this would result in the monitor watcher being exposed to 164.3 alarms, which is 20.5 alarms per hour or one alarm every 3 minutes. This may become more important as the “war room” concept of remote ECG monitoring becomes more mainstream, since it is unclear how many monitors a monitor watcher can safely watch before missing critical changes in patient condition.

Signal Quality and Abrasive Skin Preparation

The intervention of an abrasive skin preparation was tested to determine whether this confers any benefit to either signal quality or monitoring alarms. Our results did not demonstrate a significant improvement in signal quality in either electrode configuration by performing an abrasive skin preparation. There was a larger demonstrated change for the Lund than for the Mason-Likar configuration, but this failed to reach statistical significance. This research team believes that the more keratinized skin on the outer deltoids may explain the larger improvement in signal quality with abrasive skin preparation.

As with the signal quality aim of the study, the false lethal arrhythmia alarms were also not significantly affected in either the Mason-Likar or Lund electrode configuration with the abrasive skin preparation. For the skin preparation intervention in this study, there was no significant change in signal quality or false arrhythmia alarms. It is not clear from this study whether signal quality and arrhythmia alarms are associated.

In this study, an abrasive skin preparation was used prior to placing the electrodes. As was the case with others studies examining skin preparation strategies, there were limitations. Adhesive-backed abrasive sandpaper makes as the source of skin preparation makes it difficult to control for the amount of pressure applied during each of the swipes of the abrasive. Patient factors such as skin thickness, age, turgor, and reactivity to abrasion were also not measured in this study.

There are other types of skin preparation techniques suggested by other researchers; however, none of them have been examined in a continuous monitoring situation. No other skin preparation procedure was examined in this study, so the findings are limited to only a manually performed abrasive skin preparation. It is important to note that a more complex skin preparation procedure is likely to add complexity to attaching a patient to continuous monitors affecting adherence to the regimen. This study used an abrasive skin preparation in conjunction with wet-prep electrodes, not gel-backed electrodes. It is not clear from this study or a review of electrode literature whether gel-backed electrode performance would provide similar or different results in signal quality. From this research, ECG signal quality was no different on average when an abrasive skin preparation was used with wet-prep electrodes. Generalizability of these findings to gel-backed electrodes is somewhat limited due to the use of only wet-prep electrodes.

Skin Preparation on Patient Perceptions

Patients did not report any significant discomfort with the skin preparation, nor did any

subjects remove themselves from the study due to skin preparation issues. It can be assumed by no discontinuation of the study and no post-hoc complaints of skin irritation, itching, pain, redness, or other cutaneous manifestations that the abrasive skin preparation has untoward effects on these hospitalized patients. Future studies should examine other skin preparation techniques in the contexts of gel-backed and wet-prep electrodes, and various patient factors including ethnicity, pigmentation, age, and hydration status.

Patient Perceptions of Being Monitored

There appears to be no significant patient comfort difference between the Mason-Likar and Lund electrode configurations. Specific questions about bothersome electrodes were asked, but few subjects commented on any electrode location without regard to which configuration. For the four patients making a comment about electrode sites being bothersome, they all rated the bother as "mild". From this study, there was no difference in patient perceived comfort between the Mason-Likar and Lund electrode configurations.

Patients reported feeling “safer” with continuous monitoring and someone watching their heart; however, there were many patients that felt tethered and movement-limited due to the monitoring equipment. In the ICU where the monitors are at the bedside unlike the PCU in this study that had only a central monitoring station, ICU patients reported being awakened at night by the monitor with little movement. Several commented that hearing the alarms on the bedside ICU monitor made them concerned that something was changing with their heart.

Panic Attacks During the Study Period

Both of the panic attack subjects, as well as the two patients that chose to discontinue their study participation around the 8-hour mark had their records examined. All four of these patients had similarities. There were 2 men and 2 women, and all had prior psychiatric diagnoses of anxiety and/or depression. All four subjects were current tobacco users, with two of them citing the need to leave the clinical area to smoke cigarettes as the primary reason for

discontinuing the study and hospital ECG monitoring. All four of the study participants that withdrew from the study, also withdrew from clinical ECG monitoring with the hospital system. It was unclear from subject interviews whether this was related to being confined to the clinical unit or from the amount of combined monitoring equipment being used for clinical management and the research study.

Strengths and Limitations

Strengths

This is the first study to use a validated, computerized ECG signal quality algorithm to measure ECG signal quality. Compared to prior research that used 60-seconds of ECG data from 20 healthy volunteers, this study used an average of 23.8 hours of continuous ECG data for 100 patients in a real-world hospital in two electrode configurations. Synchronous recording of ECG signals represents novelty of this study from the Mason-Likar and Lund electrode configurations, so that direct comparisons can be made. Every effort was made to replicate the offline analysis of data to be the same as the hospital monitoring system.

No prior studies have examined the baseline patient perceptions of wearing a Mason-Likar configuration. Comparison of Mason-Likar to the Lund configuration regarding comfort is difficult, but this study allowed patients to make a direct comparison of comfort of each electrode site in each configuration. Addition of the qualitative data to the data collection tool strengthened this study design, by introducing another facet of monitoring, the patient perception. Patient perceptions of being monitored are absent in ECG science, so this study represents the first contribution to the use of this type of information in driving patient-centered monitoring forward.

A major strength of this study is that blinded analysis was performed using unbiased computerized algorithms without knowing skin preparation status or the electrode configuration source. This data was not revealed to blinded analysis experts in case further analysis will need to be accomplished. This use of the computerized ECG signal noise measurement is feasible

and reproducible, and should be considered for use on the publicly available databases, such as MIMIC, in the future. For arrhythmia alarm analysis, this study used the same version of software for the offline analysis as the research site hospital was using at this time. It is a limitation of the study that nursing changes to parameter alarms was not consistently obtained on every subject.

Experienced clinicians accomplished alarm adjudication for true or false alarm status with advanced training and clinical experience in arrhythmia analysis. Alarms that were difficult to determine were escalated for verification by other experienced clinicians in the Drew ECG Monitoring Research laboratory for arrhythmia determination.

General Limitations

This study was conducted at a single university medical center; however, there was a nearly even distribution by gender, and a good representation of ethnicities. Since most patients were wearing a hospital monitoring system, the addition of 2 additional Holter monitors presents some sampling artifact, especially with information about perceptions of carrying equipment. Although several researchers have used this technique in the past, this is still a factor to consider in evaluating subjective information from these patients.

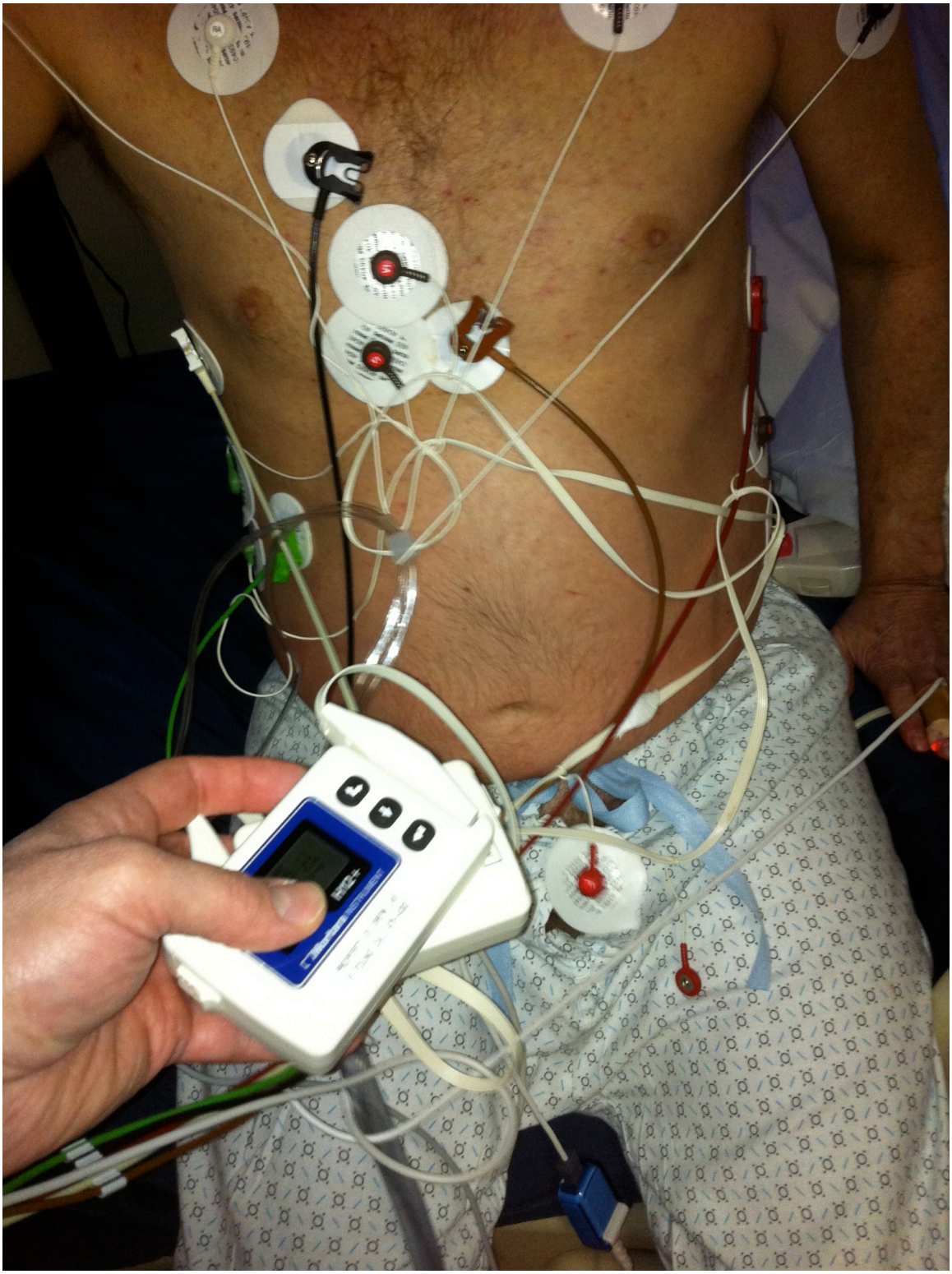
Technical limitations

This study did present a minimal to moderate amount of subject burden as demonstrated in the subject photo included as **Figure-5.1**. With study Holter monitors plus a hospital telemetry box or ICU hardwired configuration, there were a total of 12 study electrodes plus 5 hospital electrodes on every study participant. Each electrode had a corresponding wire connecting the electrodes to monitoring equipment. There is direct competition for electrode sites at the V1 and V5 positions; so exact placement using standard monitoring electrodes is technically impossible as shown in this subject at the time of study monitor removal in **Figure-5.1**. Although double-connector electrode could have been used, the overarching plan for this study was to gather

data as close to clinical practice as possible, including clinical equipment such as electrodes.

These data do not replicate the hospital monitoring environment exactly, since individual patient alarm settings were not know for the offline waveform processing with EK-Pro® to replicate hospital monitoring practice for that particular individual. It is unknown whether this would have increased or decreased the number of false lethal arrhythmia alarms. For this reason, the number of alarms analyzed and reported as a part of this study reflects the alarm burden with the monitoring system running at factory default settings. The lethal arrhythmia alarms studied are not modifiable in the clinical monitoring system, so this impact should be minimal on the numbers of alarms reported. This would have been more important in an analysis of threshold alarm parameters, but would not impact the detection of asystole, ventricular fibrillation, and ventricular tachycardia.

Figure 14-Technical Limitations of Connecting Multiple Competing Electrode Configurations at the V1 and V5 Sites as Demonstrated in this Patient.



The use of wet-prep electrodes was a conscious decision by the research team. These electrodes are not routinely used in many clinical settings, and there was no gel-backed electrode component to this study. Recommendation for future research would include a direct comparison of gel-backed and wet-prep electrodes to determine whether a difference exists in signal quality and alarm rates, both with and without a form of skin preparation. Based on the findings of this study, improving the signal quality does not appear to be an immediate solution for improving the clinical alarm fatigue issues. Signal quality may be important in other aspects of monitoring such as QT interval measurement and ST-segment analysis for ischemia; however, neither QT interval nor ST-segment alarms were studied.

In this first round of analysis, this study did not examine ST-segment alarms for ischemia monitoring, nor did the researchers examine non-lethal alarms. Published practices suggest a movement to re-categorize many non-lethal arrhythmia alarms, such as PVC-type alarms, to advisory alerts without audible alarms (Graham & Cvach, 2010). It is not clear in the current literature what impact this may have on recognition of deteriorating patient status or in-hospital cardiac arrest outcomes.

Measurement limitations

Although used for standard 10-second ECG analysis, the Hook-Up Advisor® was adapted for use in continuous monitoring. FDA-approved for the standard diagnostic 12-lead ECG, this technology appears to have face and construct validity in measuring signal quality. For future studies, the Hook-Up Advisor® appears to be a relatively simple, efficient, and reliable method for measuring signal quality, although it is proprietary software.

EK-Pro® has been used for arrhythmia analysis as a part of many GE Healthcare monitoring products. It is known that high false positive arrhythmia alarm rates are common with all monitoring equipment, and there is the notion that to have high sensitivity, we must make a sacrifice in specificity. Future research may consider using multiple tools to measure post-

marketing arrhythmia alarm frequency among various manufacturer algorithms to determine precise sensitivity and specificity performance for each.

A measurement limitation of this study is that signal quality was measured by evaluating low-frequency high amplitude baseline wander, and high-frequency low-amplitude myoelectric noise. It was assumed that the notch filter on the bedside monitoring equipment removed 60 Hz interference; however, review of signal quality data showed multiple episodes of 60 Hz interference were identified. Future analysis should examine how effective notch filters are at removing this powerline type of interference in the ECG, and how many of the alarms were associated with predominantly 60 Hz interference.

Many of the alarms that were triggered for asystole and bradycardia had low amplitude R-waves that could be obliterated in a small amount of ECG signal noise. In these cases, the patient could be in a “green” or “yellow” signal quality state and experience a false asystole or bradycardia alarm. Compared to other studies, our dataset has the limitation of having only collected physiologic signals from the ECG without invasive pressures, pulse oximetry, or other physiological variables.

Asystole and Bradycardia Alarms Could Be Reduced

Detection of R-waves is the primary step for human or automated analysis of an ECG waveform. If no R-wave is detected, no further analysis can be conducted and the conclusion is that the patient is in asystole. Electrode disconnects from lead wires and electrode dislodgement from the skin surface are common reasons for a flat ECG waveform devoid of any electrical activity. Prior work done by this research team showed that asystole alarms by both ECG and arterial line discontinuation were causes of numerous alarms (Fidler, 2011 #369). Future technology should also be able to differentiate an electrode disconnect situation from the absence of cardiac electrical activity. From a hospital practice perspective, it should be considered a routine practice to disable the alarms at the central monitoring station prior to

electively discontinuing monitoring to avoid the alarms associated with the disconnection.

Careful examination of the ECG waveforms during alarm adjudication showed that there were other common conditions triggering asystole alarms other than a disconnection. Low amplitude ECG waveforms with small R-waves made it difficult for human and computerized algorithm identification of any cardiac electrical activity. There are current minimum guidelines from the American National Standard Institute (ANSI) / Association for the Advancement of Medical Instrumentation (AAMI) for ECG equipment to detect R-waves as long as they are greater than 0.5mV in amplitude, although modern equipment has more sensitivity (ANSI/AAMI, 2007 #370). This issue needs to be addressed in a clinical context since a small R-wave less than 0.5mV is challenging to detect, but refutes the diagnosis of asystole.

A low amplitude R-wave with even a small amount of interference in the ECG signal can obliterate small to medium amplitude R-waves. In the context of the Weaver-Shannon Model of Information Theory and Nyquist's Communication Theory, there were many conditions where the noise exceeded the signal. It is conceptually easy to understand why a human or algorithm could misinterpret the ECG signal if only a single ECG lead is available for analysis of the waveform. There were multiple occasions where at least one of the five measured ECG leads had a good enough signal quality to identify R-waves and suppress a false asystole alarm. Manufacturers should consider using simultaneous multi-lead analysis as a redundant measure prior to alarming falsely for asystole. Future research needs to examine how some of the leads maintain a clean signal where others do not, or perhaps certain leads are more noise immune than others. It should be conceptualized that use of all available ECG data should be incorporated into a simultaneous multi-lead analysis as an integral part of automated ECG rhythm interpretation algorithms to avoid false asystole alarms.

Recommendations for Clinical Practice

After completing the analysis of data from this study, there are several recommendations

for clinical practice, and the section that follows recommends areas for future research. As far as ECG signal quality and the Mason-Likar and Lund configurations are concerned, the Mason-Likar is more noise immune in ambulatory subjects. For continuous clinical monitoring in the PCU, Mason-Likar is recommended over the Lund for a more noise immune signal, which is very important if ST-segment ischemia alarms are activated. In the ICU, there was no difference in signal quality between the Mason-Likar and Lund electrode configurations making either configuration an option.

It is known that when using the Mason-Likar configuration for monitoring, the addition of precordial electrodes to obtain a diagnostic 12-lead ECG using the bedside monitoring equipment is an efficient mechanism. Since there are differences between a Mason-Likar and a standard 12-lead ECG, a consensus must be reached whether the limb electrodes will be moved to the distal extremities to obtain an ECG from standard electrode sites, or when the bedside obtained ECG will be relabeled as a nonstandard recording of the ECG. Alternatively, if the Lund were accepted as a the electrode configuration, the 12-lead ECG obtained using bedside monitoring equipment would be more similar to the standard ECG at the price of decreased signal quality. Consensus will need to be reached regarding the prioritization and weighting of criteria for determining the ideal electrode configuration for continuous hospital monitoring.

Abrasive skin preparation was well tolerated by the patients in this study; however, skin preparation did not make a significant impact in either signal quality or false lethal arrhythmia alarms. It is unclear whether there would have been a difference in either signal quality or false lethal arrhythmia alarms if gel-backed electrodes were studied instead of the wet-prep electrodes that were used in this research. Future research should focus on alternate methods of skin preparation to improve signal quality when used with different types of electrodes. On a related topic, when hair is present at the electrode sites, most patients reported that they would prefer to have the site shaven to reduce pain at the time of electrode removal.

Improving the patient perception and satisfaction with being monitored should be in the foreground of clinicians and monitoring manufacturers. Many believe that alarm fatigue is a phenomenon that occurs in clinical staff, particularly nurses; however, patients are experiencing a different form of alarm fatigue. Adjusting parameter alarms to tailor the individual patient alarm profile is an effective method to reduce the burden of false alarms for patients and staff. Patients would like to have options to carry telemetry boxes, and many subjects asked why they could not wear pants with a pocket to carry monitoring equipment. Offering alternative types of electrodes for patients that develop a rash or itching with certain electrodes may alleviate an annoyance for the patient.

Monitoring manufacturers should consider patient comfort as a top priority in the next generation of monitoring equipment. Patients complained of the weight of the telemetry boxes, the intrusion of reinforced lead wires into their face when changing position, and thickness of lead wires when lying down to sleep. Creating smaller versions of monitoring equipment was a frequent request by patients to alleviate the nuisance of carrying monitoring equipment. Patients requested waterproof monitoring equipment that could be worn in the shower, and similarly, nurses requested waterproof equipment in case of accidental toilet immersion.

As described by Drew (2011), the ideal electrode configuration assessment tool does assign a weighted value to each of the elements to consider about a proposed configuration, but these weights may need to be adjusted depending on the population being monitored.

Future Research Needed

In light of these findings, it seems that the Fidler-Drew Conceptual Model for alarm fatigue and false alarms has some limitations. Hinging on noise in the ECG signal as the source for misdiagnosis and alarm fatigue, this model fails to demonstrate several components of the model. First, the skin preparation intervention did not impact the signal quality significantly. Second, the noise in the ECG signal was not associated with more false arrhythmia alarms.

Other conceptual models will need to be constructed to better represent what, if any, interactions exist between signal quality, skin preparation, and false arrhythmia alarms.

Further analysis of this dataset is already in the planning stages and will include many aspects beyond the scope of the aims of this study. Since this study was conducted in a single center in the United States, it may be of interest to determine differences in practice and equipment in a global market place. There is little research examining ethnic and racial disparities in monitoring practices, or difference that should be considered in recommending electrode placement, skin preparation, and other interventions aimed at improving signal quality and the alarm fatigue crisis.

It is not clear whether the results would be similar or different if gel-backed electrodes were used in this study. Basic science research would suggest that wet-prep electrodes, although considerably more expensive, may provide a more stable ECG signal earlier in the monitoring period than gel-backed electrodes; however, this difference in signal quality may not translate to a difference in monitoring alarms (Tam & Webster, 1977). It is also not well understood if a larger improvement in signal quality may be obtained with a skin preparation and gel-backed electrodes. The frequency of electrode changes is not well delineated in this study or other literature, and further work should aim at determining the usable lifespan of a skin electrode.

Improving the user experience in human-technology interaction design may be of importance in improving signal quality and reducing false monitoring alarms. Improved human factors interaction design between monitor and staff to improve signal quality and data entry into the computerized algorithm may be significant. As an example, an alert from the monitor at the time of admission to enter crucial information to alert the system of the presence of a pacemaker, a bundle branch block, or low amplitude QRS state may improve monitoring accuracy.

Future iterations of automated arrhythmia recognition software should consider additional user interface features to attempt to increase the amplitude of the R-wave, reduce the noise in the signal, or both. Incorporating a low-voltage alert into the admission programming for a patient may recommend strategies to improve R-wave amplitude such as shaving hair from the electrode contact points, alternate skin preparation strategies, and alternate electrode placement techniques.

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APPENDIX 1. CHR APPROVAL

Human Research Protection Program
Committee on Human Research

Notification of Expedited Review Approval

Principal Investigator / Co-Principal Investigator
Barbara J Drew Richard L Fidler CRNA, NP, MSN, MBA
Type of Submission: Continuing Review Submission Form
Study Title: Determining the Ideal Electrode Configuration for Hospital Cardiac Monitoring
IRB #: 10-04962
Reference #: 046490
Committee of Record: Laurel Heights Panel
Study Risk Assignment: Minimal
Approval Date: 06/12/2012 **Expiration Date:** 07/06/2015
Regulatory Determinations Pertaining to this Approval (if applicable):
Individual HIPAA authorization is required of all subjects.
A waiver of HIPAA Authorization is acceptable for the recruitment procedures to identify potential subjects. The recruitment procedures involve routine review of medical or other records, do not adversely affect the rights and welfare of the individuals, and pose minimal risk to their privacy, based on, at least, the presence of the following elements: (1) an adequate plan to protect the identifiers from improper use and disclosure; (2) an adequate plan to destroy the identifiers at the earliest opportunity consistent with conduct of the research, or a health or research justification for retaining the identifiers was provided or such retention is otherwise required by law; (3) adequate written assurances that the requested information will not be reused or disclosed to any other person or entity, except as required by law, for authorized oversight of the research study, or for other research for which the use or disclosure of the requested information would be permitted by the Privacy Rule;
(4) the research could not practicably be conducted without the waiver; and (5) the research could not practicably be conducted without access to and use of the requested information.
All changes to a study must receive CHR approval before they are implemented. Follow the modification request instructions. The only exception to the requirement for prior CHR review and approval is when the changes are necessary to eliminate apparent immediate hazards to the subject (45 CFR 46.103.b.4, 21 CFR 56.108.a). In such cases, report the actions taken by following these instructions.
Expiration Notice: The iMedRIS system will generate an email notification eight weeks prior to the expiration of this study's approval. However, it is your responsibility to ensure that an application for continuing review approval has been submitted by the required time. In addition, you are required to submit a study closeout report at the completion of the project.
Approved Documents: To obtain a list of documents that were approved with this submission, follow these steps: Go to My Studies and open the study – Click on Submissions History – Go to Completed Submissions – Locate this submission and click on the Details button to view a list of submitted documents and their outcomes.
For a list of all currently approved documents, follow these steps: Go to My Studies and open the study – Click on Informed Consent to obtain a list of approved consent documents and Other Study Documents for a list of other approved documents.
San Francisco Veterans Affairs Medical Center (SFVAMC): If the SFVAMC is engaged in this research, you must secure approval of the VA Research & Development Committee in addition to CHR approval and follow all applicable VA and other federal requirements. The CHR website has more information.

APPENDIX 2. CONSENT

UNIVERSITY OF CALIFORNIA, SAN FRANCISCO
CONSENT TO PARTCIPATE IN A RESEARCH STUDY

Study Title: *Determining the Ideal Electrode Configuration for Hospital Monitoring*

This is a medical research study. The study team, Dr. Barbara Drew, RN, PhD or Richard Fidler, NP, MSN from the Department of Physiological Nursing will explain this study to you. If you have any questions, you may ask the study team.

Medical research studies include only people who choose to take part. Take your time to make your decision about participating. You may discuss your decision with your family and friends and with your health care team. If you have any questions, you may ask your study doctor.

You are being asked to take part in this study because you have been admitted to the hospital and will have your heart monitored by electrocardiogram (ECG) while you are in the hospital.

Why is this study being done?

In this study, the researchers are seeking to learn more about the difference between two ways of putting the skin electrodes (stickers) on the body. Typically, the skin electrodes for the heart monitor are placed on the chest under the right and left collar bones, one on each lower ribcage, and one on the chest below the left nipple. An alternative way of placing these electrodes is to put change the position of the electrodes from under the collar bones to the outside of both arms, about half way down the upper arms. In particular, the researchers are interested in knowing more about differences between the two different ways of putting the electrodes on the body to decrease the amount of interference in the electrocardiogram, determine whether there are differences in the number of false alarms on the monitor, and then to determine whether placing the electrodes in different places either bother patients too much or become dislodged accidentally. The researchers are not paying you for this research. The researchers do not have any financial or proprietary interests to gain from conducting this study.

How many people will take part in this study?

About 100 people are planned to participate in this research study.

What will happen if I take part in this study?

- **Before you begin the main part of the study:**
    - Your medical chart will be reviewed by the study team.

- **During the main part of the study,** if you agree to be in this study, you will have two small additional monitors placed on your body. By participating in the study, in addition the monitoring electrodes that will be attached to you to guide your medical care, you will also be connected to two (2) small, lightweight Holter (heart) monitors. To compare the two different places to put electrodes, one Holter monitor will be connected to electrodes on the

chest under the collarbones, and the second Holter monitor will be connected to electrodes on the outer shoulders. This will take about five minutes.

Once the monitors are connected, they will be placed in a pouch with a neck strap so that you can move around as your medical condition permits. Each of the electrodes will have a circle drawn around each of the electrodes so that if the electrodes are accidentally dislodged or removed, they can be replaced in exactly the same position where it belongs.

- **When you are finished with the monitoring experiences,** you will be asked to point to any electrodes (if any) were bothersome. If any of them caused you a problem, the study staff will ask you to further describe the problem(s) that the electrode(s) caused. The researchers are very interested in the patient perspective for each of the electrode sites.

How long will I be in the study?

Participation in the study will take about twenty-four (24) total hours to collect the monitoring data.

Can I stop being in the study?

Yes. You can decide to stop at any time. Tell your nurse, physician, or the study staff if you are thinking about stopping or decide to stop. He or she will tell you how to stop your participation safely. The study doctor may stop you from taking part in this study at any time if he/she believes it is in your best interest, if you do not follow the study rules, or if the study is stopped.

What side effects or risks can I expect from being in the study?

The researchers see no significant risk for injury due to participation in this study; however, wearing several sets of electrodes and wires may be a nuisance to some people while they are hospitalized. To complete the study, we are asking that each person wear the study monitors for a 24 hour time period.

Strict attention to privacy and confidentiality will be maintained, but as with any research, there is a risk that privacy and confidentiality could be breached. To maintain confidentiality, your monitoring data will be assigned a study participant ID number, and all of your data will be identified only by this ID number. A separate list with your name associated with your study ID will be kept separate from the data files, and this list will be kept locked in the Drew ECG Monitoring Research Lab at UCSF. After the data is gathered and analyzed, the list of names will be destroyed to minimize the risk of confidentiality and privacy breaches.

Are there benefits to taking part in the study?

There is no direct benefit to you. The data obtained from the research Holter monitors will be analyzed later after it is disconnected from you. These Holter monitors are in addition to the heart monitoring you will receive as a part of your hospitalization. However, this study will

help the researchers learn more about monitoring the electrocardiogram, and it is hoped that this information will help in the treatment of future patients in the hospital that wear heart monitors.

What other choices do I have if I do not take part in this study?

- Getting standard treatment for your condition without being in a study.

  You do not need to enroll in this study, and not enrolling will not change the care provided to you during your hospitalization.

Will my medical information be kept confidential?

We will do our best to make sure that the personal information in your medical record is kept private. However, we cannot guarantee total privacy. Your personal information may be given out if required by law. If information from this study is published or presented at scientific meetings, your name and other personal information will not be used.

Organizations that may look at and/or copy your medical records for research, quality assurance, and data analysis include:

- UCSF Committee on Human Research

What are the costs of taking part in this study?

You will not be charged for any of the study activities.

Will I be paid for taking part in this study?

You will not be paid for taking the time to participate in this study.

What happens if I am injured because I took part in this study?

It is important that you tell the study staff, Dr. Barbara Drew or Richard Fidler, if you feel that you have been injured because of taking part in this study. You can tell the staff in person or call him or her at 415-378-4559.

Treatment and Compensation for Injury: If you are injured as a result of being in this study, treatment will be available. The costs of the treatment may be covered by the University of California depending on a number of factors. The University does not normally provide any other form of compensation for injury. For further information about this, you may call the office of the Committee on Human Research at 415-476-1814.

What are my rights if I take part in this study?

Taking part in this study is your choice. You may choose either to take part or not to take part in the study. If you decide to take part in this study, you may leave the study at any time.

No matter what decision you make, there will be no penalty to you and you will not lose any of your regular benefits. Leaving the study will not affect your medical care. You can still get your medical care from our institution. Participating or not participating in this study will have no influence on your medical management or care during your hospitalization.

We will tell you about new information or changes in the study that may affect your health or your willingness to continue in the study.

In the case of injury resulting from this study, you do not lose any of your legal rights to seek payment by signing this form.

Who can answer my questions about the study?

You can talk to the study staff about any questions, concerns, or complaints you have about this study, 24 hours per day. Contact the study staff, Richard Fidler at 415-378-4559.

If you wish to ask questions about the study or your rights as a research participant to someone other than the researchers or if you wish to voice any problems or concerns you may have about the study, please call the Office of the Committee on Human Research at 415-476-1814.

CONSENT

You have been given copies of this consent form and the Experimental Subject's Bill of Rights to keep.

You will be asked to sign a separate form authorizing access, use, creation, or disclosure of health information about you.

If you wish to be in this study, please sign below.

Date	Participant's Signature for Consent
Date	Person Obtaining Consent

APPENDIX 3. DATA COLLECTION FORMS

PRE-APPLICATION OF HOLTER MONITORS DATA

**** Please verify that the consent form has been signed by patient_____(initials)
Date of data collection: _____ (mm/dd/yyyy) **Time of attachment**:_____
Medical Record number:_____ **ICU** or **PCU Room number ___** **Ambulatory** (Y / N) ?
Holter monitor number: *ML electrode configuration* # _____ Reconfirmed by ____(initials)
*Lund electrode configuration#*_____ Reconfirmed by ____(initials)
Holter cable number: *Have you circumscribed the ML electrodes with RED marking pen?* (Y / N)
Have you circumscribed the Lund electrodes with PURPLE marking pen? (Y / N)
Have you marked each of the electrodes at the 12 and 3 o'clock positions? (Y / N)
Age: _____ in years **Gender:** male/female
Height: _____ inches **Weight**: _____pounds
Ethnicity: *African-American* *Asian* *Caucasian* *Latino (white or black)* *Mixed*
Vital Signs at time of attachment: *HR*_____bpm *BP* _____/_____
*RR*_____ Temp_____
Admitting Dx:_____ **Date of admission**: _____ (mm/dd/yyyy)
Typical Medical problems: (circle all that apply)

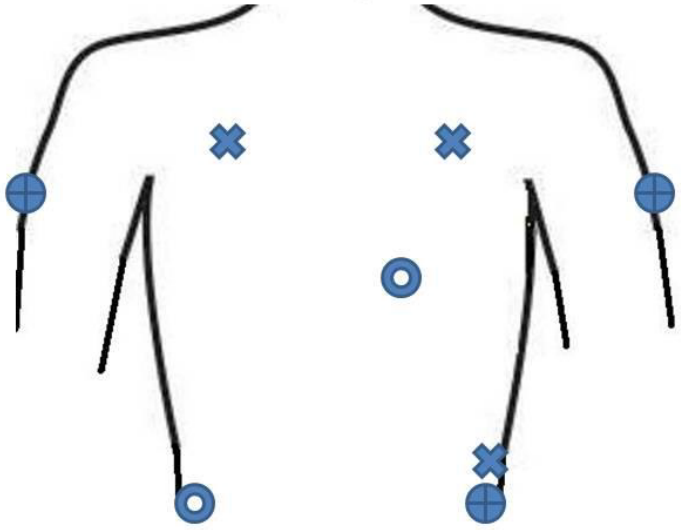
HTN	Anxiety/depression	Prior Cardiac Arrest
DM	Liver disease	Known CAD
Tobacco	Renal disease	Arrhythmia specify (_____)
Cocaine/Other drugs (_____)	Pacemaker/ICD	
Hyperlipidemia	CHF	Other_____

Lab Data: please record date and time of lab data
ABG: pH_____pO2_____pCO2_____HCO3_____SaO2_____iCalcium_____
CBC: WBC____Hgb_____Hct_____plt_____
Chem: Na_____K_____Cl_____CO2_____BUN_____creat_____
ECG data: (computerized interpretation) date ____/___/_____ and time ____:_____
Ventricular Rate_____PR_____QRS_____QT_____QTc_____rhythm_____QRS axis_____T wave axis_____
CXR data: (radiologist interpretation) date ___/____/_____ time____:_____
Report_____
_____
ECHO data: EF _____ LVH (yes/no)_____E/A reversal? (y/n) Diastolic Dysfunction (y/n)

HOLTER MONITOR REMOVAL DATA

ICU or **PCU bed number _____ OOB chair? _____Ambulatory** (Y / N) ?
Were the Holter monitors removed prior to the end of the study? (Y / N) **Time were they removed? _____**
Reason for removal:
_____
Date / time scheduled to remove the Holter monitors: ___ /____/_____(mm/dd/yyyy)
Time : _____
Are the ML and Lund Holters are still connected to the correct electrode configuration? (Y / N)
How many of the 5 ML electrodes are the original adhesive electrodes: _____
Which of the ML electrodes are intact at 24 hours: *LA RA LL ground V1 V5*
How many of the 5 Lund electrodes are the original adhesive electrodes: _____
Which of the Lund electrodes are intact at 24 hours: *LA RA LL ground V1 V5*
Please indicate all (ML or Lund) electrode(s) with skin irritation.
_____
If communicative, ask the subject "Were any of the electrode sites bothersome?" (Y / N)
Ask the subject to point to the bothersome electrode site and rate the bother 1-10, *where 1 is minimal bother and 10 is a major problem.*

Figure 1. Mason-Likar compared to Lund Electrode Configurations



Mason-Likar electrode positions–
LA and RA midclavicular line below clavicle, LL below left ribs on lateral abdomen
Lund electrode positions—
LA and RA are on lateral deltoids at level of axillary folds, LL on left lateral hip
Shared sites for both configurations– right lower abdomen (ground), precordial unipolar chest (V) lead
** *The difference between configurations is the LA, RA, and LL positions only.*

Removal Data Part II

Laboratory Data (include all including time)

Chemistry

CBC

Abg

CXR data (if any)
Echo data (if done during the data collection)
Events (date and time)

Arrhythmia

Ischemia

Procedures
Pre Attachment Monitor Alarm settings **Post Attachment Settings**

APPENDIX 4. SUMMARY OF HOOK-UP® ADVISOR SIGNAL QUALITY STATISTICS

sfstudy
cg\drew\twentyfourhourtest.bin

r	nsegs	nevents	gl	gll	gV1	gV5	yl	yll	yV1	yV5	rl	rll	rV1	rV5	gTot	yTot	rTot	
0	360	32	315	308	313	313	45	52	47	47	0	0	0	0	278	82	0	
1	360	53	277	274	279	277	82	85	80	82	1	1	1	1	238	121	1	
2	360	58	258	261	264	266	102	99	96	94	0	0	0	0	222	138	0	
3	360	53	261	255	263	250	99	105	97	110	0	0	0	0	205	155	0	
4	360	45	286	298	292	294	73	61	67	65	1	1	1	1	262	97	1	
5	360	72	236	226	226	229	120	130	130	127	4	4	4	4	159	197	4	
6	360	70	217	208	222	209	143	152	138	151	0	0	0	0	137	223	0	
7	360	64	247	259	244	247	113	101	116	113	0	0	0	0	174	186	0	
8	360	82	226	243	238	233	134	117	122	127	0	0	0	0	154	206	0	
9	360	71	269	248	250	250	90	111	109	109	1	1	1	1	163	196	1	
10	360	60	258	273	267	265	101	86	93	95	1	1	0	0	221	138	1	
11	360	28	311	315	309	313	49	45	51	47	0	0	0	0	273	87	0	
12	360	35	300	311	302	298	60	49	58	62	0	0	0	0	262	98	0	
13	360	26	326	329	327	337	32	29	31	21	2	2	2	2	308	50	2	
14	360	8	353	355	356	354	7	5	4	6	0	0	0	0	352	8	0	
15	360	4	354	356	357	356	6	4	3	4	0	0	0	0	353	7	0	
16	360	18	346	346	349	348	14	14	11	12	0	0	0	0	339	21	0	
17	360	15	342	349	348	345	18	11	12	15	0	0	0	0	339	21	0	
18	360	3	358	358	357	357	2	2	3	3	0	0	0	0	357	3	0	
19	360	35	322	322	320	315	38	38	40	45	0	0	0	0	298	62	0	
20	360	51	305	312	305	305	54	47	54	55	1	1	1	0	278	81	1	
21	360	52	274	278	264	283	83	79	93	74	3	3	3	3	238	119	3	
22	360	75	271	271	270	269	86	86	87	88	3	3	3	3	219	138	3	
23	360	75	211	227	226	227	149	133	134	133	0	0	0	0	134	226	0	
4	8640	1085	6923	6982	6948	6940	1700	1641	1676	1685	17	17	16	15	5963	2660	17	

APPENDIX 5. SAMPLE HOOK-UP ADVISOR® EVENTS OUTPUT

Event log: AC : BLW : MA : EN

00:00:10 : Yellow : .... : .... : x... : .... : Start of event. Green duration: 00:00:10

00:00:20 : Green. Event duration: 00:00:10

00:00:30 : Yellow : .... : .... : x... : .... : Start of event. Green duration: 00:00:10

00:00:40 : Green. Event duration: 00:00:10

00:01:00 : Yellow : .... : .... : .x.. : .... : Start of event. Green duration: 00:00:20

00:01:10 : Green. Event duration: 00:00:10

00:02:00 : Yellow : .... : .... : x... : .... : Start of event. Green duration: 00:00:50

00:02:10 : Yellow : .... : .... : xx.. : ....

00:02:20 : Green. Event duration: 00:00:20

00:02:40 : Yellow : .... : .... : xx.. : .... : Start of event. Green duration: 00:00:20

00:02:50 : Green. Event duration: 00:00:10

00:04:00 : Yellow : .... : .... : .x.. : .... : Start of event. Green duration: 00:01:10

00:04:10 : Yellow : .... : .... : x... : ....

00:04:20 : Yellow : .... : .... : xx.. : ....

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6/13/2014
Date